THE EFFECTS OF BACKPACK WEIGHT ON THE BIOMECHANICS OF LOAD CARRIAGE

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An analysis of the effects of 4 backpack loads (6, 20, 33, and 44 kg) on walking gait was performed on 16 male volunteers using a cinematographic system, force platform, tri-axial accelerometer, and 6 surface electrodes located over the trapezius, spinal erector, quadricep, hamstring, gastrocnemius and tibialis anterior muscles. When the load became very heavy, stride frequency increased. Double-support as percent of stride increased along with the load, effected by a delayed floor push-off. Knee range of motion increased with load during the eccentric knee flexion period from heel-strike until mid-stance. A lower total body center of mass position as the load increased was effected both by greater knee flexion and a more forward leaning trunk. An initial propulsive impulse at heel-strike resulted from flexion at the knee rather than from extension at the hip. A protective gait adjustment when increasing to the heaviest load limited the medial travel of the center of mass. As the load increased, hip extensor torque increased proportionately. Yet knee extensor torque increased more than expected, while ankle plantarflexor torque increased less than expected. Trapezius muscle activity showed that the frame-and-belt system did not prevent the shoulders from supporting considerable load. The spinal erectors produced the largest burst of activity at contralateral heel-strike. The gastrocnemius was largely inactive except for high activity during push-off, which did not increase with very heavy loads. The burden of carrying a very heavy load fell less on the calf muscles than on the muscles around the knee and hip. Trunk forward/downward excursion and acceleration increased with load. The erector spinae acted eccentrically to decelerate trunk motion as the trunk approached its maximum forward lean. Slack in the straps enabled peak forward acceleration of the pack to occur later and be of lower magnitude than the peak forward acceleration of the trunk. However, a similar effect did not occur in regard to peak backward backpack acceleration. Concentric/eccentric resistance exercises that strengthen the quadriceps, spinal erectors, and abdominal muscles may help improve load carriage performance. Backpacks must be designed to distribute a major portion of the load to the hips.

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BACKGROUND

While there has been much research on the biomechanics of human gait, only a small proportion of such research has specifically addressed load carriage. In 1981, Pierrynowski, Norman, and Winter used cinematography to investigate variation in the mechanical energy levels of the body segments and efficiency of volunteers carrying five different backpack loads. Kinoshita and Bates (43) compared the effects on ground reaction forces of a standard backpack vs. a two-pack system, the latter of which distributed the load equally between the front and back of the volunteers. In another study, Kinoshita (44) reported significant changes from unloaded body posture and gait pattern when loads of 20% and 40% of body weight were carried, but less deviation from normal walking with a front/rear pack system than a standard backpack. Our laboratory compared the effects of a load carriage system that distributed the load between the front and back of the torso to the effects of a standard backpack on walking posture both before and after a fatiguing maximal speed 20 km road march (26, 36). We also compared various load carriage systems as to walking and running biomechanics among both male and female soldiers (33, 34). Electromyography has been used to evaluate muscle activity during walking, especially in the lower extremities (8, 11, 49). Yet most studies of load carriage have been physiological rather than biomechanical and have focused on metabolic response (4, 17, 23, 31, 35, 51, 65).

Many investigators have biomechanically analyzed unloaded human locomotion, using methodology that can be applied to the study of load carriage. They evaluated stride length (70, 72), joint forces and moments (9, 13, 38), joint ranges of motion (50), path of the center of pressure on the foot (32, 77), mechanical power (47, 79), external work (28), timing of gait events (77), braking impulse (54), and the effects of speed on mechanics (64).

Electromyography (EMG) can help determine which muscles are involved in a physical activity, estimate their contraction intensity, and determine the muscle contraction sequence (5, 18, 67, 68). Stulen and De Luca (75) used EMG frequency analysis to gain insight into the effects of fatigue on motor unit recruitment patterns.

Most of the commercial and military backpack systems and other load carriage equipment available today have not been tested biomechanically. Application of quantitative biomechanical evaluation to loaded human locomotion can potentially contribute to the effectiveness of equipment evaluation and design. Thus, the purpose of this study was to gather information on the effects of backpack load on gait kinematics and kinetics in order to form the basis of recommendations concerning pack systems, physical training programs, and load carriage technique. Ultimately, this could benefit people who engage in load carriage for whatever purpose by increasing load capacity and transport speed, lessening the likelihood of injury, improving efficiency, and decreasing perceived level of difficulty.

During walking, at least one foot is always in contact with the ground, while in running, there is a flight phase during which neither foot is in contact with the ground. Distance variables like stride length, step width, and step length, and temporal variables

like stride and step duration, cadence, and speed provide additional quantitative information about gait. Some of these variables may be affected by factors such as age and sex.

In order to determine muscle activity during the various phases of the walking cycle, electromyography (EMG) has been used in conjunction with force platform and cinematographic analysis (82). The EMG record tells which muscle is acting during the gait cycle and can provide some measure of how much force the muscle is generating.

In 1984, a biomechanics research program was established at USARIEM. Because of its relevance to the Army, load carriage was selected as a major area of focus of the program. The study described in this report was undertaken to increase knowledge about the effects of the weight carried on the kinematics and kinetics of gait, and on the pattern and degree of muscle involvement revealed by electromyography.

LIST OF SYMBOLS, ABBREVIATIONS, AND ACRONYMS

mph	miles per hour
USARIEM	U.S. Army Research Institute of Environmental Medicine
USASBCC	U.S. Army Soldier, Biological, and Chemical Command
USASSC	U.S. Army Soldier Systems Center in Natick, MA
ALICE	All purpose, lightweight, individual carrying equipment

EXECUTIVE SUMMARY

In order to gather information on the effects of backpack weight on gait biomechanics, an analysis of load carriage gait using a cinematographic system, a force platform, a tri-axial accelerometer, and six surface muscle electrodes was performed. Sixteen male volunteers walked with packs of 6, 20, 33, and 47 kg while electrical activity of the trapezius, spinal erector, quadriceps, hamstrings, gastrocnemius, and tibialis anterior muscles were monitored. When the load became very heavy, stride frequency increased, probably helping reduce mechanical stress to the bones of the legs and feet, including the metatarsal bones, which may be especially susceptible to stress fractures during load carriage. Double-support as percentage of stride increased along with the load, effected by a delayed foot push-off, especially when going from the next heaviest to the heaviest load, improving control and stability. Knee range of motion increased with load during the eccentric knee flexion period from heel-strike until mid-stance, helping reduce heel-strike shock. A lower total body center of mass position as the load increased, effected both by greater knee flexion and a more forward leaning trunk, helped control the potentially destabilizing effect of the load, reducing moment of inertia about the feet. The initial propulsive impulse at heel-strike resulted from flexion at the knee rather than from extension at the hip, and was effected by hamstring muscle activity. A lack of increase in medial ground reaction force between the next-heaviest and heaviest loads indicates a protective gait adjustment limiting the medial travel of the center of mass, possibly limiting forces experienced by the small muscles that adduct and control abduction of the hip. As the load increased, hip extensor torque increased proportionately. However, knee extensor torque increased more than expected, while ankle plantarflexor torque increased less than expected. As evidenced by trapezius muscle activity, the frame-and-belt pack did not prevent the shoulders from supporting considerable load. Though the trunk inclined forward as load increased to keep the pack-plus-body center of mass over the feet, the adjustment did not bring the center of mass as far forward over the foot as without a load. The spinal erectors produced their largest burst of activity at contralateral heel-strike. The load carriage stride was characterized by concentric knee flexion at heel-strike, eccentric knee flexion during a shock absorption phase, concentric knee extension during push-off, and a quiescent period after toe-off during the swing phase. The stride was also characterized by eccentric tibialis anterior activity at heel-strike, which controlled the rate of foot plantarflexion, which did not increase with load. The gastrocnemius was largely inactive except for high activity during push-off, which did not increase with very heavy loads. The burden of carrying a very heavy load fell less on the calf muscles than on the muscles around the knee and hip. Trunk forward/downward excursion and acceleration increased with load. The erector spinae acted eccentrically to decelerate trunk motion as the trunk approached its maximum forward lean. Slack in the straps enabled peak forward acceleration of the pack to occur later and be of lower magnitude than the peak forward acceleration of the trunk. However, a similar effect did not occur in regard to peak backward backpack acceleration of the trunk. Concentric/eccentric resistance exercises that strengthen the quadriceps, spinal erectors, and abdominal muscles may help improve load carriage performance. Backpacks must be designed to effectively distribute a major portion of the load to the hips.

DEFINITIONS OF TERMS

For readers interested in the biomechanics of load carriage, but unfamiliar with its terminology, the definitions below will be helpful for understanding this report:

1. **Stride Time**. The time for a full stride, which includes both a left and a right step. We measure stride time as the time between consecutive right heel-strikes.

2. **Stride Length**. The length of a full stride, which includes both a left and a right step. We measure stride length as the horizontal distance between the locations of two consecutive right heel-strikes.

3. **Stance Phase**. When a given foot is in contact with the ground. It begins with the foot's heel-strike and ends with its toe-off. Each complete gait cycle includes a stance phase for each foot. The stance phase makes up about 60% of the walking gait cycle with little variation for age and height at normal walking speed (55, 71).

4. **Swing Phase**. When a foot is not in contact with the ground. It begins with the foot's toe-off, continues as the foot swings forward, and ends with its heel-strike. Each complete gait cycle includes a swing phase for each foot. The swing phase makes up about 40% of the walking gait cycle, with little variation for age and height at normal walking speed (55, 71).

5. **Single-support**. The period during a gait cycle when only one foot is in contact with the ground; i.e., one foot is in its stance phase while the other foot is in its swing phase. A single-support period of the right foot begins at toe-off of the left foot and ends at the subsequent heel-strike of the left foot. Each complete gait cycle includes a single support phase on each foot.

6. **Double-support Phase**. The period during a gait cycle when both feet are in contact with the ground at the same time; i.e. both feet are their respective stance phases. Each complete gait cycle includes two double-support phases. One begins as the right heel strikes the ground while the left foot is still on the ground. It continues as weight is shifted from the left foot to the right foot and ends when the toe of the left foot leaves the ground. The other begins as the left heel strikes the ground while the right foot is still on the ground. It continues as weight is shifted from the right foot to the left foot and ends when the toe of the right foot leaves the ground.

7. **Ground Reaction Force**. The force exerted by the ground on the foot, which is equal in magnitude and opposite in direction to the force exerted by the foot on the ground.

8. **Joint Torque**. The net impetus exerted by the muscles around a joint to rotate adjacent body segments towards or away from each other around the joint; it is quantified as the muscle force times the perpendicular distance from the line of action of the force to the pivot point of the joint.

9. **Impulse**. The area under the curve of force as a function of time.

10. **Kinematics**. Quantification of motion without regard for the forces producing the motion. Human kinematic data include linear and rotational position, velocity, acceleration, and range of motion for each body segment and the total body center of mass. It also includes such variables as stride length, stride frequency, and relative time in single- and double-support.

11. **Kinetics**. Analysis of the forces and torques that bring about motion. Human kinetic data include ground reaction forces, joint bone-on-bone forces, and muscle torques.

12. **Electromyography**. Recording and analysis of muscle electrical activity.

INTRODUCTION

KINEMATIC ASPECTS OF GAIT

Studies of human gait are inherently complex because of the interrelationships among the various parameters that describe walking and running. For example, the carriage of a load while walking is a factor that affects gait pattern (29, 37, 43, 44). Both the high and low placement of the load on the back cause forward body lean, bringing the knees, hips, shoulders and head further forward (7). A double-pack, which distributes the load between bags on both the front and back of the torso, produces less forward lean of the trunk than does a backpack (36, 44). In addition, the double-pack is associated with greater stride frequency and shorter stride length than is the backpack (36), which is regarded as a positive adaptation. However, the double-pack did not reduce the effects of fatigue on loaded walking posture (26). Female soldiers differed somewhat from male soldiers in their biomechanical response to carrying loads (59). Under load, the females moved their knees through a more limited range of motion, bending them more but straightening them less. They also evidenced less average forward trunk lean, but a greater range of trunk motion, and also spent a greater percentage of the stride in double-support. Yet load carriage efficiency, as measured by oxygen consumption per unit of body-plus-load mass, did not differ between the sexes.

Effects of Backpack Loads

While it has been shown that stance duration doesn't change significantly with increasing backpack load, swing duration significantly decreases as the load increases to up to 50% of body weight (29, 44, 52, 60). This results in an increased percentage of stride in double-support as the load increases (37). A load on the back during walking, equivalent to 20% of body weight, resulted in a decrease in maximum knee flexion during the swing phase (29). Our own studies (33, 34) showed several effects of increased load on gait biomechanics including increases in ground reaction forces, shoulder strap pressure, double-support duration, and trunk range of motion, but decreases in knee range of motion.

Kinoshita (44) investigated the effects of different loads (double-pack and backpack) on selected biomechanical parameters of walking gait. Ten healthy males were tested at a walking speed of 1.25 meters per second with external loads of 0%, 20% (light), and 40% (heavy) of body weight. Body-segment orientation and joint angles revealed that during the initial weight-bearing phase, there was greater knee flexion accompanied by lesser hip extension for the heavy load condition. The loads caused significant forward lean of the trunk, which averaged about 11 degrees. It was also noted that greater dorsiflexion occurred during the early mid-support phase as the load increased. The foot rotated anterioposteriorly around the distal end of the metatarsal bones for a longer period of time when the heavy load was carried. The author concluded that to take stress off the metatarsal bones, step length should be shortened as the load is increased so that a faster transfer of the body weight from one leg to the other can be accomplished. Our own study of front-back load carriage systems vs. standard backpacks (26, 36), in which loads of 34,

48, and 61 kg were carried, showed considerably less forward inclination of the trunk with the front-back pack than the backpack, and the effect increased with the load.

Martin and Nelson (52) studied 11 males and 11 females walking at a speed of 1.78 meters per second with five loads (0, 9, 17, 29, and 36 kg). The 9 kg load consisted of a military utility shirt, gym shorts, military boots, and web gear. For the 17 kg load, a helmet and armor vest were added. Weights were placed in a framed rucksack to achieve the highest two loads. The males and females showed significantly different gait patterns under all load conditions. The females evidenced a higher rate of stepping than the males and a corresponding shorter stride length. The walking patterns of both male and female volunteers were affected by the increases in load carried. There was an increased forward inclination of the trunk but only for the two heaviest loads carried in the rucksack. Findings on stride length and stride rate were mixed, with these variables showing either no increase or significant increase as load increased. No significant changes occurred in men or women in stance time, but swing time decreased consistently with loading. Alterations in double-support percentage were small and inconsistent.

Pascoe et al. studied ten children, 11-13 years of age, and found that when they carried a 7.7 kg backpack load (17% of body weight), their stride length significantly decreased and their stride frequency significantly increased compared to a no load condition. Carrying the backpack load also brought about a significant forward head and trunk lean.

KINETIC ASPECTS OF GAIT

Ground Reaction Forces

To kinetically analyze performances in which two parts of the body come into contact with an external object (which may include the ground), it is necessary to directly measure the force exerted by at least one of those body parts on the external object. This applies to activities such as walking, manual labor, and load carriage. The necessary information cannot be inferred from movement studies only, using methodology such as goniometry or cinematography.

During running, no more than one foot makes contact with the ground at any given time. Thus it is possible to calculate forces and torques on the body from kinematic data and knowledge of the volunteer's body mass. However, during walking, both feet contact the ground at the same time during the two double-support phases of each full stride. Therefore, force platform data are required to enable a kinetic analysis. As the foot exerts force on the ground during the stance phase of a stride, the ground exerts equal and opposite force on the foot. The study of ground reaction forces during walking can provide relevant information about the mechanics of gait under various conditions. It provides a direct measure of impact forces on the foot, and thus is relevant to the understanding and prevention of lower extremity injuries.

Force platforms, which use sensing elements whose electrical characteristics change in proportion to the magnitude of applied forces, are used to measure the forces and moments applied by the foot on the ground. If a complete force and torque record of a footstep is to be obtained, each of the force and moment components must be sampled at a sufficiently high rate. An example of the use of force platform technology is the diagnosis of hip joint problems through evaluation of the vertical component of ground reaction force during walking, decomposition of the force into the sum of its harmonic components, and description of the force in other mathematical terms (41). Bresler and Frankel (13) studied different characteristics of vertical ground reaction force measured on a force platform. Yamashita and Katoh (83) used a specially designed force platform to analyze the pattern of center of pressure during level walking.

Schneider and Chao (66) analyzed the ground reaction forces of 26 normal volunteers during walking. The curve of vertical ground reaction force as a function of time typically had a dual-hump shape with the second peak higher (114% of body weight) than the first (106% of body weight). When graphed as a function of time, vertical ground reaction force formed a pattern that was nearly symmetrical about a vertical line at 50% of the stance phase of each foot. The front-back ground reaction force was not symmetrically distributed, with a larger peak forward (propulsive) force (19.0% of body weight) and a smaller peak rearward (braking) force (15% of body weight). The waveform of the medio-lateral ground reaction force was more irregular than that of the other two ground reaction forces. The medial ground reaction force was predominant except at both ends of the stance phase.

Several ground reaction forces increase with weight of load carried including those in the downward, forward, rearward, and medial directions (37, 44). Lateral ground reaction forces are not as clearly affected by load (37).

Kinoshita and Bates (43) conducted a study of ground reaction forces using five male volunteers walking with five different loads at walking speeds within a range of 1.17 to 1.33 meter per second. The vertical force curves were all bimodal and similar for all five conditions. During the no-load carrying condition, the first and second peaks respectively occurred on average at 20% and 75% of stride starting from heel contact. The minimum vertical force occurred at 45%. While the first peak and minimum point between the peaks remained constant, the second peak occurred later as the load was increased. First peak, second peak, and minimum force values and the total vertical impulse were proportional to the increase in the system weight. There was a tendency towards similar adjustment to load seen in the anterior-posterior force curves. The maximum braking and propulsive forces also increased proportionally to the increase in the system weight.

In another study of the effects of the load carrying system (44), ground reaction force curves showed differential effects for two different carrying systems. Peak anterior-posterior force and minimum vertical force occurred later with a front-rear double-pack than with a normal backpack. Maximum braking force was lower, and minimum vertical force was higher for the double-pack than the backpack. The author suggested the differences

between load carriage systems were largely attributable to their differential effects on body posture during walking.

Our own load carriage studies (33, 34) showed the effects of load on ground reaction forces. It was found that peak heel-strike and peak push-off vertical and braking ground reaction forces all increased with the load carried.

Joint Moments And Forces

Understanding the effects of forces on material bodies is essential to the study of locomotion. The strength of a rotational impetus is called moment of force and is equal to the magnitude of the force multiplied by the perpendicular distance from the line of action of the force to the point of rotation. A kinetic analysis of walking (2, 41, 66) and running (50, 54, 81) revealed basic patterns of moments generated by the muscles around the ankle, knee, and hip. However, individual differences in pattern of moments about the knee and hip during gait have also been noted (61, 82).

Simon et al. (69) investigated the forces generated at heel-strike during human gait using both a force platform and a force transducer inserted into the heel of the shoe. The output traces were analyzed for the existence of high frequency impulsive loads during a normal walking cycle. The data showed that during normal human gait the lower limb is subjected to a high impulsive load at heel-strike. The severity of this impulse varied with the individual, the walking velocity, the angle with which the limb approached the ground, and the compliance of the two materials coming in contact at heel-strike. Peak force varied from 0.5 to 1.25 times body weight and its frequency components varied from 10 to 75 Hz.

Several other studies (50, 54, 81) revealed basic moment patterns during running. Winter (81) studied ankle, knee, and hip moments while 11 normal volunteers jogging at slow speed. He found that the moment of force for the total lower limb was primarily extensor during the stance phase. He also noted the relative timing of the peak extensor torques at the three joints. The hip peaked at 20% of stance, the knee at 40% of stance, and the ankle near 60% of stance. The variability of the moment patterns across all jogging trials was considerably less than that seen during walking. Two power bursts were seen at the ankle, including an absorption phase early in the stance followed by a dominant generation peak during late push-off. Average peak power generation was 800 W with individual maximums exceeding 1500 W.

ELECTROMYOGRAPHIC ANALYSIS OF GAIT

Over the years electromyography (EMG) has been used to investigate the activity of the muscles of the lower extremity during walking. It provides a recording of muscle electrical activity between two conducting electrodes, which vary in type and construction. The two main types of EMG electrodes are surface electrodes and indwelling (needle and wire) electrodes. Each has its advantages and its disadvantages.

The needle electrode is the most common type of indwelling electrode used for clinical diagnostic purposes but is unsuited for studies of movement. One of two main advantages of the needle electrode is that its small pickup area enables the electrode to detect individual motor unit action potentials during low-force contractions. The other advantage is that it may be repositioned within the muscle so that new territories may be explored (6). It can pick up EMG signals from muscle fibers up to 1.5 mm away. The needle electrode can be used to detect signals from deep muscles, and receive signals from a much more confined area than surface electrodes. In order to obtain EMG signals from an entire muscle group, a number of electrodes would have to be used, which would reduce the volunteer's comfort.

The fine wire electrode is another type of indwelling electrode. It is extremely fine and easily implanted and withdrawn, therefore painless. The main purpose of using wire electrodes in human movement studies is to record a signal that is proportional to the contraction force of a muscle. A limitation of the wire electrode is its tendency to migrate after it has been inserted during the first few contractions of the muscle. Basmajian and De Luca (6) suggested that muscle with the electrode be contracted and relaxed at least one-half dozen times before any measurements are taken.

Surface electrodes may be used effectively with superficial muscles but, because they pick up signals from a broad area of muscle near the skin, cannot be used to detect signals from small, deep muscles. The main advantage of surface electrodes is that they are convenient to use and provide high fidelity EMG signals. Surface electrodes are acceptable when the time of activation, frequency, and magnitude of EMG signals are to be examined, but small and/or deep muscles are not the objects of interest.

Some studies have been undertaken which used electromyography to examine neural control of gait (12, 22, 61). The results of those studies provided some indication of when certain muscles are on and off during the gait cycle but have not given quantitative measures of the intensity of muscle activation. There are significant changes in EMG timing and magnitude as walking speed changes (45).

During level walking, the hamstrings and tibialis anterior reach peak activity at heel-strike. Quadriceps muscle activity increases thereafter to keep the knee from buckling and then to push off. The hamstrings and quadriceps show elevated EMG activity starting just before and continuing until just after toe-off (48). The calf muscles increase their activity gradually from the mid-stance phase until toe-off. The knee stabilizing function of the gastrocnemius is most important during the stance phase (76). The calf muscles are active during knee extension and ankle dorsiflexion during the mid-stance phase (27). Even after quadriceps activity ceases during the mid-stance phase, knee extension continues, due to torque about the knee resulting from movement of the upper body center of mass forward of the knee joint (55).

Effects of Backpack Loads on Muscle Electrical Activity

In several studies, muscle activity patterns of the leg and back muscles were examined during walking with loads (8, 19, 29, 37). An electromyographic study (19) of the lumbar paraspinal muscles during load carriage was undertaken with a group of 24 healthy volunteers (12 males and 12 females). Four different carrying positions (i.e., contralateral, ipsilateral, anterior or posterior) and two different loads (10% and 20% of body weight) were compared to walking without loads during the stance phase of each gait cycle at a speed of 1.3 m/s. The results showed significant effects of load. Compared to walking without an external load, lumbar paraspinal EMG activity showed slight decrease when loads were carried.

Under heavier backpack loads the spinal erectors are clearly more active than during unloaded walking. Harman et al. (37) found that spinal erector EMG activity decreased for relatively low loads (less than 33 kg) but increased sharply when loads reached 47 kg. While the gastrocnemius muscles showed increases in EMG activity with load, results from the trapezius and quadriceps muscles were mixed, showing either no increase or significant increase with load. Load did not significantly affect EMG activity of the tibialis anterior and hamstrings.

Norman, Winter, and Pierrynowski (56) investigated muscle activities of the rectus femoris, gastrocnemius, lumbar erector spinae, and trapezius among six male volunteers carrying loads of 0, 15, 19, 23, 29, and 34 kg (20-40% of body mass). There were no significant differences in rectus femoris activity due to loads. Erector spinae EMG activity was higher at the 29 and 34 kg loads than for the no load condition and activity at the 34 kg load was higher than at all other loads. Trapezius EMG activity at the 34 kg load was higher than at the zero and 15 kg loads.

Placement and type of backpack affect muscle activity. Bobet and Norman (8) studied the effects of two different load placements (just below mid-back or just above shoulder level) on muscle activities of erector spinae and trapezius on 11 volunteers. The volunteers walked on a smooth level surface at an average velocity of 5.6 km/hour (1.56 m/sec) carrying a load of 19.5 kg in a specially designed backpack. Both muscle activities significantly decreased with mid-back placement. In an experimental study on the effects of pack design, Holewijn (39) compared a backpack with frame and hip belt to a frameless backpack. The backpack with frame and hip belt produced lower EMG activity in the trapezius muscle than did the frameless backpack because the former transferred support of the load from the shoulders to the hips.

Holewijn (39) monitored the EMG signal of the trapezius pars descendens muscle among four young male volunteers as they walked on a treadmill carrying either 0, 5.4, or 10.4 kg in a backpack. The load significantly increased the root mean square EMG value of the muscle, corresponding to an increase in force exerted by the muscle.

PHYSIOLOGICAL CORRELATES OF GAIT MECHANICS

The physiological responses of volunteers carrying loads, especially as to energy cost, have been examined in some detail. Energy cost increases in a systematic manner with increases in body weight (24, 30), load (3, 10, 21, 46, 73, 74), velocity (74), and grade (10, 30, 58). It has been reported that the natural walking cadence is most efficient (40).

Mechanical analysis of walking has been studied for several years (20). Cavagna and Margaria (16) introduced energy calculations from force platform data, with the body regarded as a point mass. Winter, Quanbury, and Reimer (79) developed a mechanical energy calculation method based on a segment-by-segment analysis assuming energy exchanges within segments and energy transfer between adjacent segments.

Martin (51) evaluated the effect of lower extremity loading on measures of mechanical work done on the lower extremity, rate of oxygen consumption ($\dot{V}O_2$), and heart rate. Five load conditions (no added load, and loads of 0.50 kg and 1.00 kg added to either the thighs or feet) were examined. $\dot{V}O_2$ and heart rate increased as load increased on both thighs and feet. The increases in $\dot{V}O_2$ due to foot loading were nearly twice as great as those due to thigh loading. The results also demonstrated that 1.00 kg added to the feet produced small but significant increases in stride length and swing time, and a decrease in peak ankle velocity. Significant increases in the mechanical work done on the leg were produced by the loading.

Balogun (3) tested ten physically fit male students who carried external loads of 11.6, 16.1, and 20.6 kg. The author measured heart rate, pulmonary ventilation, oxygen consumption, ventilation equivalent, and oxygen pulse. All of the variables except ventilation equivalent changed significantly as load increased.

It appears that the most efficient way to carry a load is as close as possible to the center of mass of the body (73, 78). Datta and Ramanathan (21) studied seven men carrying 30 kg over a 1-km level distance at a walking speed of 1.4 m/sec. Loads were carried in seven different modes: in a front/back double-pack, rucksack, and rice bags on a yoke, Sherpa frame, on the head, and in the hands. The researchers concluded that the double-pack is most economical and least stressful because of the large area of muscular and skeletal support, the position of the load near the center of gravity, and freedom of the hands to both carry small items and maintain balance.

An internal-frame backpack, by its nature, is carried closer to the body than an external-frame backpack. Our study comparing both types of frames (35) showed that the internal-frame backpack, when used with its belt, produced the lowest rate of oxygen consumption for soldiers carrying a 34 kg backpack at 1.34, 1.56, and 1.79 m/s. Because we hypothesized that pack center of mass location was likely responsible, we ran a follow-up study (57) using a specially designed backpack in which the center of mass could be systematically varied. Results showed that soldiers exhibited lower oxygen consumption when the center of mass of the backpack was higher and closer to the soldier's back than

when it was lower and further from the soldier's back. The effect is likely due to the fact that a higher, closer load results in less perturbation of unloaded walking posture to get the load over the base of support, while the low, distant load requires greater forward trunk inclination.

A much less efficient way to carry a load is on either the feet or ankles. A load carried on the feet costs 5-7 times more energy than an equivalent load carried on the torso (46, 74). Stated another way, for each 0.1 kg added to the foot, the energy cost of locomotion increases 0.7 to 1% (15, 42, 46, 73). The energy cost of carrying loads on the ankles exceeds that of carrying loads in the hands close to the body by 5-6 times (73). When vigorous arm movements are involved, the energy cost of hand carriage can exceed that of ankle carriage (53). Ralston and Lukin (63) examined the effects on a single volunteer of adding load to the feet as to both mechanical work done on the body and the energy cost of walking. A 31% increase in energy cost produced by the addition of 2 kg to each foot was accompanied by a 35% increase in mechanical work. Their results suggested that the increases in the mechanical work done on the body were primarily limited to work done on the loaded segments.

Legg and Mahanty (46) compared the following five load carriage modes as to cardiorespiratory, metabolic, and subjective responses: 1) total load in a backpack with frame, 2) total load in a backpack with no frame, 3) half the load in a framed backpack and half in pouches attached to a waist belt, 4) half the load in a framed backpack and half in a front pack on the chest 5) total load in a jacket with weights inserted in pockets evenly distributed about the trunk. They reported no significant differences in the mean cardiorespiratory and metabolic costs associated with each of the five modes of load carriage. Volunteers rated the front/back pack combination and the trunk jacket more comfortable than the other load carriage modes.

PURPOSE OF THE PRESENT STUDY

The study described in this report was undertaken to increase knowledge about the effects of the load carried in a backpack. While previous studies produced relevant information about the biomechanics of load carriage, many questions remain. This study incorporates a greater range of backpack loads than in most previous studies. Our use of both cinematographic analysis and electromyography provides the opportunity to calculate body movement kinematics and the joint torques generated by muscle groups needed to effect the observed load carriage body movements, and to verify which muscles are active and to what degree. The military relevance of this study is heightened by the fact that most of the volunteers were soldiers and the backpack frame used was from the standard Army ALICE backpack, used by U.S. soldiers for decades.

METHODS

VOLUNTEERS

Testing occurred at the biomechanics laboratory of the U.S. Army Research Institute of Environmental Medicine, Natick MA. Volunteers for the experiment included permanent party military test volunteers assigned to the U.S. Army Natick Soldier Center, soldiers recruited for temporary duty as test volunteers, and military and civilian employees of the U.S. Army Research Institute of Environmental Medicine. A total of 16 volunteers were tested. Each volunteer participated in the study for a maximum of two test conditions per day.

Sample Size Estimation

A nomogram for repeated measures (14) was used to estimate the sample size. The nomogram (Appendix A) shows the minimum difference between the dependent variable means of the experimental groups, in standard deviation units, that can be found statistically significant, given a Type I error rate of 5%. The nomogram has two vertical scales, sample size on the left side and inter-trial correlation coefficients on the right side, with a diagonal scale between them representing the minimum detectable mean difference (effect size). To find the number of volunteers needed, a line is drawn from the inter-trial correlation coefficient through the desired effect size to the sample size scale. For a given effect size, the higher the inter-trial correlation coefficient, the fewer test volunteers were needed. A higher inter-trial correlation coefficient enables the researcher to detect a smaller mean difference with the same sample size.

Because the inter-trial correlation coefficients of most dependent variables analyzed in the biomechanical study of load carriage were available from pilot study, sample size estimation could be performed easily. For example, an inter-trial correlation coefficient of about 0.90 for stride length and stride frequency with effect size of 0.5 gives sample size of smaller than five. An inter-trial correlation coefficient of about 0.70 for the EMGs with effect size of 0.5 gives a sample size of 10. The inter-trial correlation coefficients for most of the variables examined were higher than 0.60. The nomogram showed that for an inter-trial correlation coefficient of 0.60 with a moderate effect size of 0.5 and a two-tailed alpha level of 0.05, 13 volunteers were needed. It was decided to test 16 volunteers in order to provide for data lost by equipment malfunction or volunteers who might terminate testing prematurely.

INSTRUMENTATION

Force Platform System

Information needed for the kinetic analysis of load carriage includes the forces exerted by the ground on the feet (ground reaction forces). A force platform provides the needed information because the ground reaction forces are equal in magnitude to and opposite in direction from the forces exerted by the feet on the force platform. Information

provided by the force platform includes the magnitudes of forces exerted by the feet in the vertical, front-back, and left-right directions relative to the walker as well as the location on the platform of the foot center of pressure. Knowledge of the latter is essential in order to calculate the moment about the ankle joint due to ground reaction force, which is directly proportional to the distance from the point of application of the force to the joint. In addition, error in calculation of torque about the ankle results in errors in torque calculations for the knee and hip, since calculations are performed in sequence from the ankle up.

A model LG6-1-1 force platform from Advanced Mechanical Technology Incorporated (Newton, MA) was used in conjunction with a model SGA6-3 amplifier designed for use with computerized data acquisition systems. The plate, which measures .61 by 1.22 m (2 by 4 feet), was mounted on a steel frame to keep it rigid and isolated from external vibrations that might cause spurious output signals. The no-damage limits of the platform were 2,200 pounds (9,800 N) of vertical load applied anywhere on the top surface or 1,200 pounds (6,700 N) of horizontal load applied perpendicular to any of the platform's sides. The system was designed to emit voltage signals proportional to forces and torques exerted on the plate's surface, which include forces in the vertical, front-back and left-right directions and torques around orthogonal axes through the center of the plate oriented in the latter three directions. Center of pressure can be calculated from the forces and torques, as specified in the AMTI force platform manual (1). The force platform and walking surfaces were made flush by building a wooden platform around the force platform. The SGA6-3 amplifier system contained a six-channel amplifier with switch-selectable gains of 1000, 2000, and 4000 for each channel. Each channel also had a selectable low-pass filter with a 10 Hz or 1,050 Hz cutoff frequency and selectable precision bridge excitation voltages of 2.5, 5, or 10.

Accelerometer

A model EGAXT3-84-c-100 tri-axial accelerometer (Entran Devices, Fairfield, NJ) was mounted in the pack during load carriage. It emitted voltage signals proportional to pack acceleration in three orthogonal directions. This temperature compensated strain gauge accelerometer measured accelerations in the range of ± 100 g in the vertical, left-right, and front-back directions. Built-in over-ranging protection prevented damage to the device. Because of a very high resonant frequency of 1,700 Hz, the accelerometer did not distort the accelerations characteristics of human movement.

Cinematography System

Cinematography has been used for several years for the biomechanical analyses of gait (7, 44, 51, 62). The process involves filming human movement with one or more cameras driven by spring, battery, or line power, at a frame rate fast enough to capture the movement with adequate resolution. In contrast to video systems, 16 mm film requires a considerable amount of light, especially indoors and at high frame rates. After the film was processed, it was projected frame by frame onto a digitizing table, where the experimenter used a pointing device to locate major joint centers of the body. The digitizing device sent table coordinates of the joint locations to a computer with which it was interfaced. Computer

programs then processed the coordinate information to calculate kinematic variables that included body segment positions, velocities, and accelerations. The volunteer's body mass and data from a force platform were processed along with the kinematic data to produce kinetic information, which included the forces and torques at each body joint.

Video analysis has supplanted cinematography to a large extent, mainly because digitizing can be accomplished automatically, eliminating the slow and tedious process of hand-digitizing film images. Video also has the advantages of immediate availability of collected data and the low cost and reusability of videotape. However, current video systems cannot rival the image resolution of 16 mm film.

One LOCAM II camera from Redlake Corp. (Morgan Hill, CA) was used to film the volunteers during load carriage. The camera can be set at precise frame rates up to 500/sec. A frame rate of 60 Hz was used for this experiment because it was fast enough to capture the body movements of interest. A faster frame rate would unnecessarily require more film and more time spent in film digitizing. The camera incorporates a timing light which places markers on the edge of the film every .01 sec to allow checking of film speed. A model 12-0101 battery pack permitted use of the camera away from AC power outlets. Model 9003-0001 floodlights (1000 watts) from Colortran (Burbank, CA) and model 18001 Mini-Mac photoflood lamps (1000 watts) from Bardwell & McAlister (Hollywood, CA) provided illumination.

For analysis, developed films were projected with an M-16C projection head from Vanguard Instrument Corp. (Melville, NY) onto an ACT23 digitizing table from Altek Corporation (Silver Spring, MD). The projector allowed one frame of the film to be seen at a time. Specific frames could be referenced using a digital frame counter. The digitizing table had a resolution of .01 mm and was connected via its controller to a model 486-33 IBM-PC compatible computer from Club American Technology Inc. (Fremont, CA).

Electromyography System

"Utah" model surface electrodes with integral preamplifiers and band pass filtering systems from Motion Control Inc. (Salt Lake City, UT) were used to record muscle potentials from the shoulder, back and legs. Each electrode was factory calibrated, with individual gains ranging from 340 to 380. Although the gain was slightly affected by the frequency of the signal being amplified, the variation in gain for signals between 60 and 500 Hz was within 2% of the range. The bandwidth of the preamplifier was 8 Hz to 33 KHz. The high input impedance of the electrodes made it unnecessary to abrade the skin or use electro-conductive jelly.

Computerized Data Collection System

The data were sent to a model 486-33 IBM-PC compatible computer from Club American Technology Inc. (Fremont, CA), including six output signals from the force

platform, three from the accelerometer, six from the muscle EMG electrodes, and one from the event marker, for a total of 15. The signals were fed into a model DAP1200/2 data acquisition and analog-to-digital converter board from Microstar Laboratories Inc. (Redmond, WA) mounted in an expansion slot in the computer. The DAP combined analog data acquisition hardware with a 16-bit microprocessor and a real-time multitasking operating system. It had 16 channels, each of which could be specified in software as single-ended or differential.

The inputs to the DAP were voltages, which the board converted to numbers. The board could perform computations on the resulting numbers before the information was sent to the computer, making data processing very fast. The gain factor was independently software selectable for each channel, with possible values of 1, 10, 100, and 1,000. Allowable voltage input ranges with unity gain were 0 to 5 V, -2.5 to +2.5 V, -5 to +5 V, and -10 to +10 V. Maximum sampling rate was 50,000 per second. The sampling rate for this experiment was 1,000 Hz for all the channels except for the EMGs. Two logical channels operating at 1,000 Hz each were used for each EMG hardware channel, so that the actual sampling rate was 2,000 Hz per EMG channel.

Backpack

A backpack (Figure 1) was specially designed for the experiment, using a standard U.S. Army ALICE external pack frame as a base. Two metal shelves were added to the frame. On the bottom shelf was mounted a metal box containing the accelerometer, a terminal for the EMG electrodes, and a junction for a multi-conductor cable through which output data could be sent to the analog-to-digital converter board mounted in the computer.

The top shelf of the pack was designed to hold weights so that the intended experimental loads could be carried in the pack. The weights were in the form of lead bricks and rectangular iron plates.

An effort was made to match as closely as possible the location of the center of mass of the experimental pack and an ordinary backpack. A pack loaded in standard fashion was balanced on a straight edge to locate its center of mass. The weights were then arranged on the experimental pack in such a manner as to match the center of mass location of the standard pack. Blocks of stiff foam were used as spacers on the shelf under the weights to make sure all of the pack loads had the same center of mass.

Two tape markers were placed on the side of the experimental pack so that the pack's position could be determined throughout a filmed trial by digitizing. The location of the actual pack center of mass relative to the markers was measured and recorded for use by the film analysis computer program.

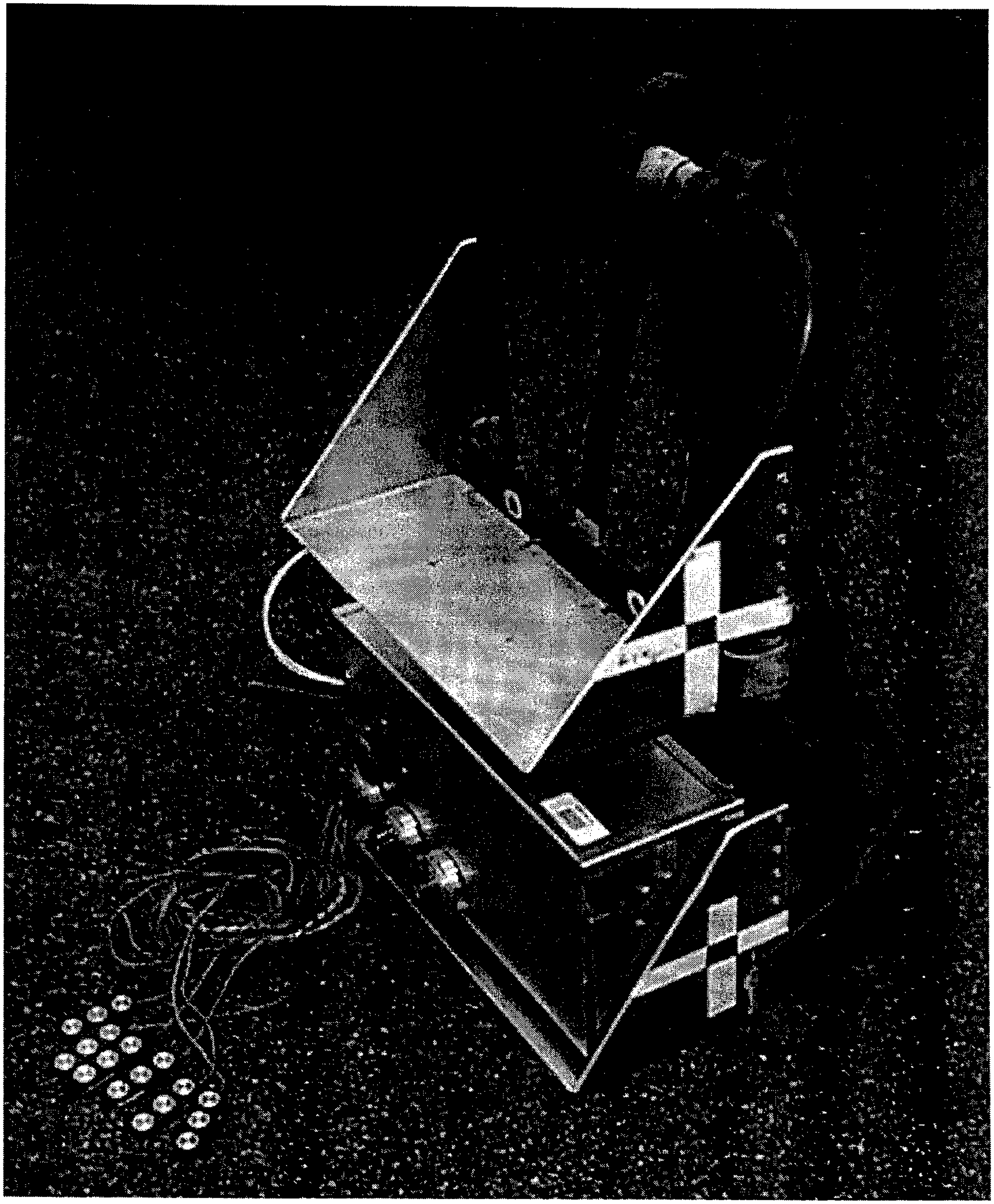


Figure 1. The experimental backpack.

Speed Cuing Device

A device to pace the volunteer's walking speed was designed at the U.S. Army Research Institute of Environmental Medicine and fabricated at the U.S. Army Soldier Systems Center in Natick, MA. It was based on a motor-driven cord marked with alternating light and dark bands that traveled around two pulley-wheels spaced 8 m apart. The speed of the cord was set using a dial. A digital speed display enabled cord speed to be set to the nearest 0.01 m/s. During an experimental trial, the device was oriented alongside the volunteer so that the visible part of the cord traveled in the direction the volunteer walked. The volunteer walked straight ahead while maintaining a peripheral view of the moving cord, which cued the appropriate walking speed.

EXPERIMENTAL PROCEDURES

Independent Variable

Load Carried. The experiment was conducted with loads of 6, 20, 33, and 47 kg. The load of 6 kg was chosen because it was the weight of the backpack itself without any additional weight on it. The volunteers had to carry the pack even in the lightest load condition because the pack contained an EMG terminal as well as an accelerometer. The load of 47 kg was selected as the upper limit of what serious backpackers and soldiers generally carry. The other two loads were selected to be equally spaced between the 6 and 47 kg loads.

Dependent Variables

The following variables were calculated from the vertical, front-back and left-right forces exerted by the feet on the force platform:

a. heel-strike and push-off peak forces (N)
b. time of occurrence of heel-strike and push-off peak force (percent of stride time)
c. peak and average front-back and mediolateral forces (N)
d. positive and negative vertical, front-back and mediolateral impulse per stride (N·sec)

Film analysis allowed calculation of the following:

a. joint ranges of motion for the hip, knee, and ankle (radians)
b. joint torques for the hip, knee, and ankle (N·m)
c. joint forces at the hip, knee, and ankle (N)
d. stride length (m)
e. stride frequency (strides/min)
f. single-support time (percent of stride time)
g. double-support time (percent of stride time)
h. body segment and center of mass position, velocity and acceleration

EMG analysis allowed calculation of the following:

a. peak and average muscle activities for the trapezius, spinal erector, quadriceps, hamstrings, gastrocnemius, and tibialis anterior muscles (uV)
b. timing of activation for the muscles listed above

Accelerometer data analysis allowed calculation of the following:

a. peak accelerations of the backpack in the vertical, front-back, and left-right directions (g)
b. timing and directions of the accelerations

Test Trials

All volunteers were orally briefed on the purpose, risks, and benefits of the study, after which they signed informed consent documents. Electrodes were attached to the volunteers' skin with adhesive tape after the skin was cleaned but not abraded with rubbing alcohol and a gauze pad. Electrodes were placed over the following muscles using anatomical landmarks according to the recommendations for standardized electrode positions (84):

- trapezius (elevates the shoulders, resists shoulder depression under the weight of the backpack)
- lower erector spinae, L4/L5 level (extends the back, resists forward movement of the trunk due to backpack weight and inertia)
- rectus femoris (extends the knee and flexes the hip during locomotion, helps lift the weight of body and backpack during the stride)
- biceps femoris (flexes the knee, extends the hip)
- tibialis anterior (works eccentrically to control the speed of foot plantarflexion so that the foot doesn't slap the ground too quickly)
- gastrocnemius (plantarflexes the foot, helps lift the weight of body and backpack during the stride)

The volunteers performed their test trials (Figure 2) while wearing shorts and military boots. Prior to data collection, reflective tape markers were placed on the right side-view joint centers of the ball of the foot, ankle, knee, hip, shoulder, elbow, and wrist. Volunteers then donned the loaded backpack. Trials consisted of walks of no more than 15 m across the force platform in the camera field of view. Each volunteer was given practice trials to adjust walking speed and starting position so that the right foot landed squarely on the force platform as the volunteer walked across it. Data for the EMGs, force platform, and accelerometer were collected for every trial, but only the data from acceptable trials were saved. A volunteer performed no more than nine trials in a test session (1 load x three speeds x three trials), with a maximum of two test sessions per volunteer per day (one in the morning and one in the afternoon). The volunteers were to walk at 1.1, 1.3, and 1.5 m/s corresponding to slow, medium, and fast walking, visually cued by the specially designed speed-cueing device running alongside the volunteer. However, later cinematographic analysis revealed that their actual speeds were respectively 1.17, 1.33, and 1.50 m/s, which

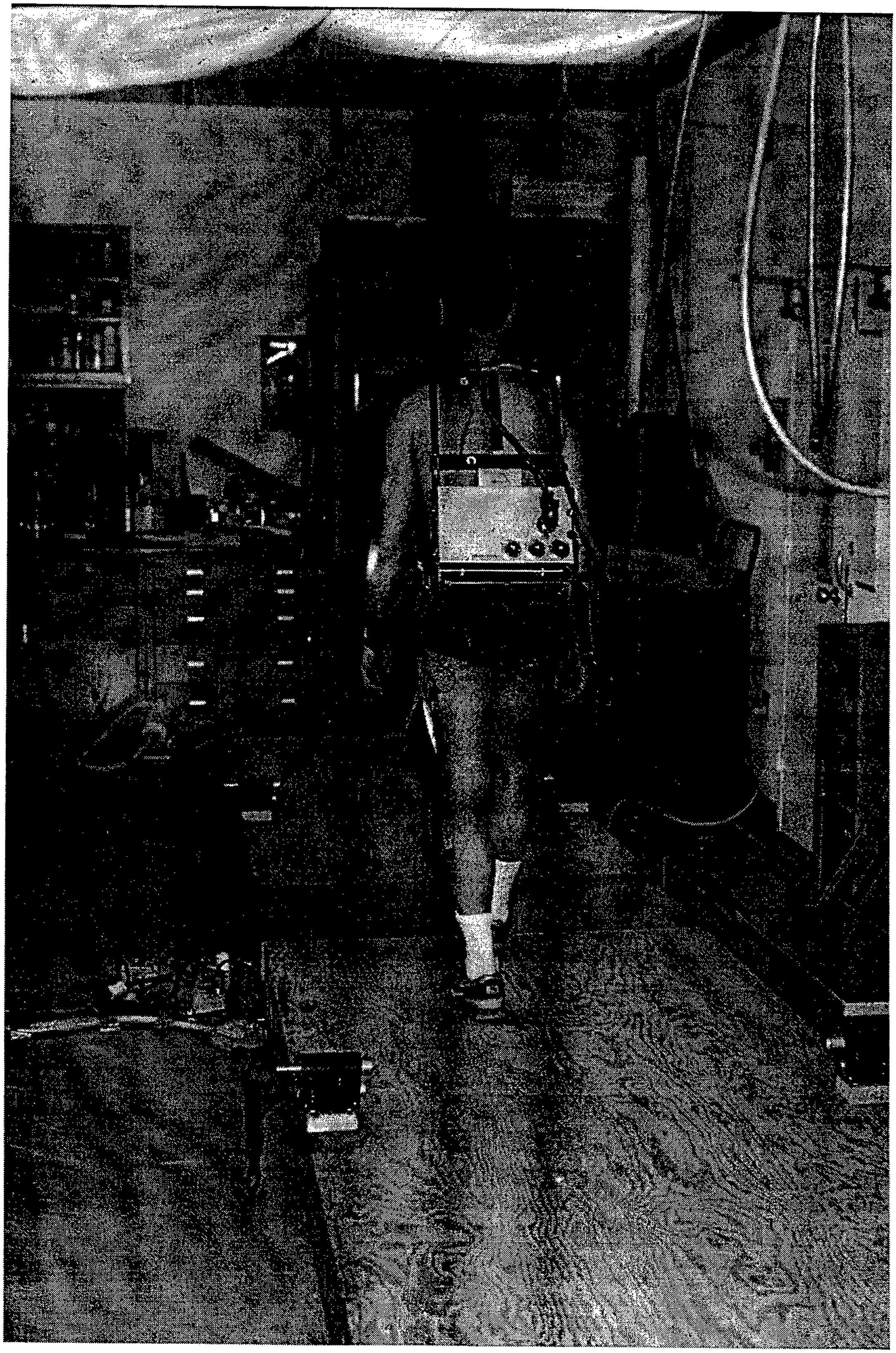


Figure 2. The experimental setup. For the actual trials volunteers wore military boots.

still can be characterized as slow, medium and fast marching speeds. Subsequent to this experiment, an electric-eye speed trap system was added to the experimental methodology to provide immediate feedback as to whether the volunteer walked at the cued speed. Each volunteer carried a different load on each test day resulting in a total of 36 acceptable trials over four test sessions. Occasionally, a trial had to be repeated if the volunteer's foot did not land directly on the force platform. Adequate rest periods were allowed between trials to avoid fatigue as a confounding factor. Each trial lasted no more than 15 seconds, so total exercise time per day was minimal.

Data Processing

Data were collected and analyzed on the computer. Programs in the C++ computer language, specifically written for the study collected the digitizing table coordinates from each frame of film, as well as the data from the six force platform channels, the three accelerometer channels, and the six EMG electrodes, all converted from analog signals to numerical information by the A/D board. Other programs performed the processing necessary to compute records of dependent variable values over the stride. A large statistical file then was created which contained key variables describing the gait patterns of all the volunteers.

The EMG data underwent digital-to-RMS conversion (67) and other interpretive procedures. The vertical and horizontal forces determined from the force platform divided by the weight of body-plus-load gave vertical, mediolateral and front-back accelerations of the system center of mass. Mathematical integration of the accelerations yielded velocities.

Digitizing. The film sequence of the load carriage trial closest to the target walking speed for the particular combination of load and walking speed was selected. An experimenter obtained the x-y image coordinates of each marker on a volunteer's body over a full stride by a process called digitizing. That process involved projecting the film one frame at a time on the rear side of the translucent digitizing table. The experimenter sequentially placed the cross-hairs of a transparent mouse-like device over the center of each joint marker image. When the experimenter pressed a button on the device, the x-y digitizer table coordinates of the marker were sent to the computer. A custom-written Borland C++ computer program collected film data from the digitizing table via an IEEE-488 interface board (Capital Equipment Corp., Burlington, MA) installed in one of the computer's expansion slots. The program drew a stick figure of the volunteer on the computer screen as the film was digitized to allow immediate detection and correction of gross digitizing errors. The computer displayed the name of each joint as it was to be digitized. If a digitizing error was made, the program allowed the user to go back and re-digitize any point at will. The program allowed a user to stop digitizing at any time, shut down the computer, and resume again at any time.

The ball of the foot, ankle, knee, hip, shoulder, elbow, wrist, and ear lobe of the right side of the volunteer were digitized. The first frame digitized was 11 frames before the frame at which the right heel passed the back of the left lower leg. The last frame digitized was 12 frames after the right heel again passed the back of the left lower leg. Because the

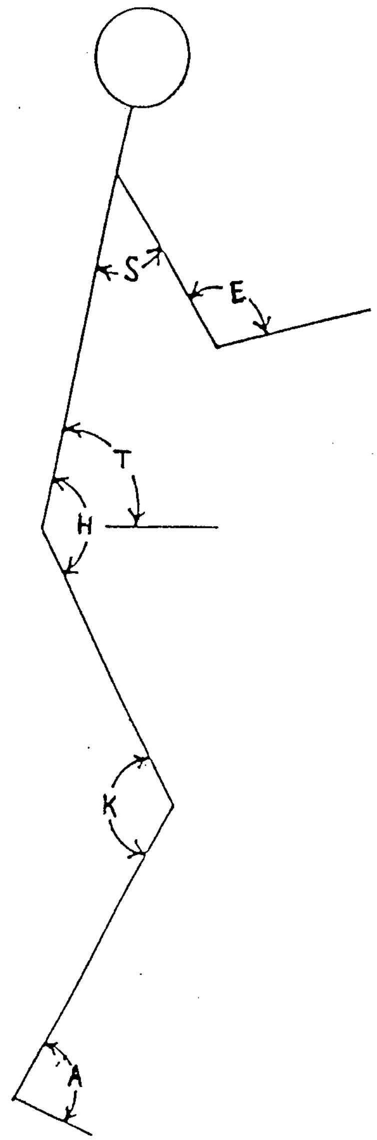
camera was aimed across the center of the force platform, this provided the best film images of a full stride. The extra frames digitized at the beginning and end of the stride were needed for mathematical data smoothing and to ensure that a full stride was recorded. At the beginning of processing the film data from each trial, the four corners of the force platform were digitized in order to later be able to calculate the film coordinates of the center of pressure.

Data Smoothing and Interpolation. The digitized film data were smoothed using Fourier analysis and Digital Filtering subroutines contained in Software for Science and Engineering Tools IPC-TC-006 (Quinn-Curtis, Needham, MA). The smoothed data were then processed with a cubic spline curve-fitting subroutine from the same software library to produce 101 interpolated frames for one full stride representing 0% to 100% of the time of a full stride. Thus, the results for each volunteer were in terms of percentage of stride. The actual time between interpolated frames was unique to each trial and was later used to calculate actual velocities and accelerations of the body segments and center of mass.

The mass, center of mass, and moment of inertia of each body segment were estimated using tables of standard body proportions based on dissection of cadavers (80). Because both heel-strike and toe-off were visible in the films and on the display of force-platform data, these two points were used to time-synchronize film and force-platform data. The EMG and accelerometer data were already time-synchronized with the force-platform data because they all were concurrently digitized by the computer's analog-to-digital converter board. The foot's center of pressure location on the force-platform's surface was calculated for each trial from force platform data using equations provided by the force-platform's manufacturer (AMTI model LG6-1-1 Biomechanics Platform Instruction Manual, 1985). Joint moments and forces for the lower extremity were calculated using segment-by-segment kinetic analysis (80).

System of Postural Analysis. To analyze posture throughout the stride, the system of body angles shown in Figure 3 was used.

Statistical Analysis. The large statistical file containing the key variables describing the gait patterns of all the volunteers was transferred to a VAX 780 main-frame computer where programs from BMDP (Berkeley, CA) were used for statistical comparisons between the different experimental conditions. An analysis of variance with repeated measures was performed on each of the variables using the BMDP 2V program. Means and standard deviations for each variable under each testing condition were calculated. An analysis with four levels of load was performed. Post-Hoc Tukey tests were employed to locate the differences between treatment means when significant treatment effects were found by analysis of variance.



A = Ankle angle: the absolute ventral angle between foot and shank. Because the foot segment endpoints were the lateral malleolus and ball of the foot, when the bottom surface of the foot was at 90° relative to the shank, the ankle angle was about 120°.
K = Knee angle: the absolute dorsal angle between shank and thigh.
H = Hip angle: the absolute ventral angle between thigh and trunk.
T = Trunk angle: the ventral angle between the trunk and a horizontal line.
E = Elbow angle: the absolute ventral angle between upper arm and forearm.
S = Shoulder angle: the angle between upper arm and trunk (positive with the upper arm in front of the trunk and negative with the upper arm behind the trunk).

Figure 3. The system of body angles used to analyze posture throughout the stride.

RESULTS

TEST VOLUNTEER CHARACTERISTICS

The test volunteers were all physically fit males, a bit above average in both height and body mass (Table 1). All engaged in regular physical activity. Of the 16 volunteers, 11 were enlisted U.S. Army personnel, three were Army officers, and two were civilian employees of the U.S. Army Research Institute of Environmental Medicine.

Table 1. Physical characteristics of the test volunteers (means±SD)

Age (yr)	30.3±9.2
Height (cm)	181.2±7.5
Body mass (kg)	76.8±8.9
Gender	all male
n	16

LOAD EFFECTS

Stride Parameters

As shown in Table 2, there were significant ($p<0.05$) load effects for all stride parameters except stride length. The post-hoc tests indicated that stride time at the 47 kg load was significantly less and stride frequency significantly greater than at the other three loads, the means of which did not differ significantly. Percentage of stride at toe-off and percentage of double-support at the 47 kg load was significantly greater than those at 6 and 20 kg, with no other significant differences for these variables.

Table 2. Stride parameters at loads of 6, 20, 33, and 47 kg (mean±SD)

Variables	6 Kg	20 Kg	33 Kg	47 Kg	Prob.
Stride time (sec)	1.21±0.09	1.20±0.09	1.21±0.09	1.18±0.08†‡§	0.000*
Stride length (m)	1.60±0.12	1.59±0.11	1.60±0.12	1.57±0.11	0.124
Stride frequency (strides/min)	49.9±3.55	50.3±3.61	50.0±3.77	51.2±3.47†‡§	0.000*
Percentage of stride at toe-off (%)	63.8±1.82	64.2±2.05	64.6±1.46	65.3±1.67†‡	0.000*
Percentage of double-support (%)	27.5±3.65	28.4±4.10	29.1±2.91	30.5±3.35†‡	0.000*

* statistically significant ($p<0.05$) load effect - ANOVA
†significantly ($p<0.05$) different from 6 kg - Tukey post-hoc test
‡ significantly ($p<0.05$) different from 20 kg - Tukey post-hoc test
§ significantly ($p<0.05$) different from 33 kg - Tukey post-hoc test

Lower Body Sagittal Plane Ranges of Motion

Figures 4 and 5 show that, across loads, minimum ankle angle occurred just before heel-strike of the opposite foot, while maximum knee angle occurred just after toe-off. There was a tendency for the greatest ankle dorsiflexion to occur during the early mid-stance phase. However, neither minimum nor maximum ankle angle nor ankle range of motion was significantly affected by test load (Table 3). Load significantly affected both minimum and maximum knee angle. Minimum knee angle at the 6-kg load was significantly smaller than at the 33 and 47 kg loads, with no other significant inter-load differences. Maximum knee angle did not differ significantly among the first three loads, but at 33 kg was significantly greater than at 47 kg. Minimum hip angle decreased significantly and maximum hip angle increased significantly as load increased, with significant differences among all loads. Maximum hip angle decreased with increasing load and was significantly different among all loads except for 20 and 33 kg. Hip range of motion increased with load, with significant differences between 6 and 33 kg, 6 and 47 kg, and between 20 and 47 kg.

Table 3. Lower body sagittal plane ranges of motion at loads of 6, 20, 33, and 47 kg (mean±SD)

Variables	6 Kg	20 Kg	33 Kg	47 Kg	Prob.
Minimum ankle angle (rad)	1.89±0.06	1.89±0.08	1.88±0.09	1.89±0.08	0.948
Maximum ankle angle (rad)	2.39±0.08	2.39±0.10	2.41±0.10	2.42±0.08	0.212
Ankle range of motion (rad)	0.51±0.07	0.50±0.05	0.53±0.06	0.53±0.05	0.079
Minimum knee angle (rad)	1.93±0.06	1.95±0.07	1.97±0.08†	1.96±0.08†	0.001*
Maximum knee angle (rad)	3.10±0.09	3.10±0.08	3.12±0.10	3.09±0.10§	0.039*
Knee range of motion (rad)	1.17±0.10	1.15±0.09	1.15±0.10	1.13±0.11	0.058
Minimum hip angle (rad)	2.52±0.12	2.44±0.12†	2.39±0.12†‡	2.33±0.11†‡§	0.000*
Maximum hip angle (rad)	3.40±0.10	3.35±0.11†	3.33±0.11†	3.28±0.11†‡§	0.000*0
Hip range of motion (rad)	0.88±0.08	0.91±0.08	0.94±0.09†	0.95±0.09†‡	.000*

Note - The joint angles are defined in Diagram.
* statistically significant ($p<0.05$) load effect - ANOVA
† significantly ($p<0.05$) different from 6 kg - Tukey post-hoc test
‡ significantly ($p<0.05$) different from 20 kg - Tukey post-hoc test
§ significantly ($p<0.05$) different from 33 kg - Tukey post-hoc test

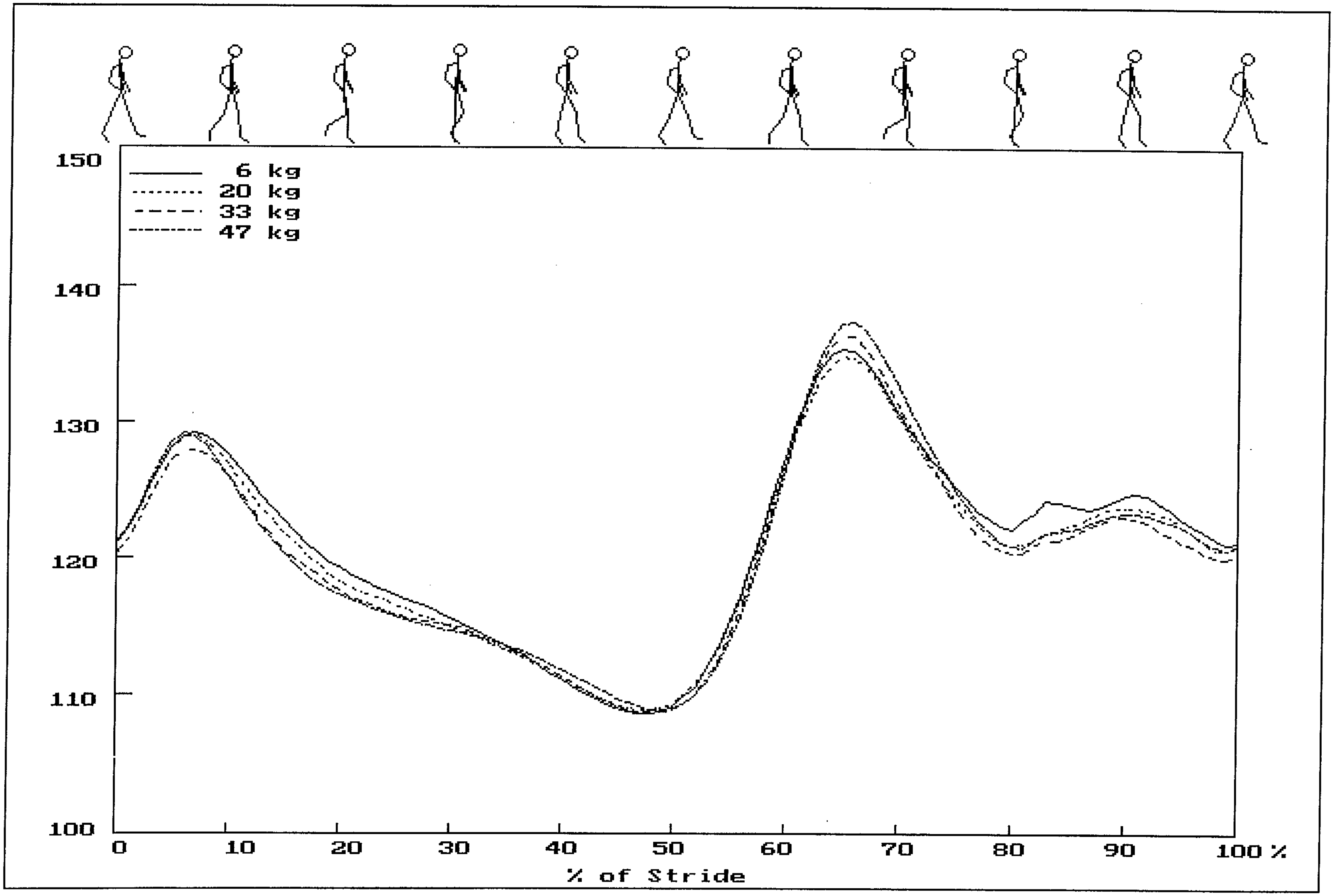


Figure 4. Load effects for ankle angle (degrees).

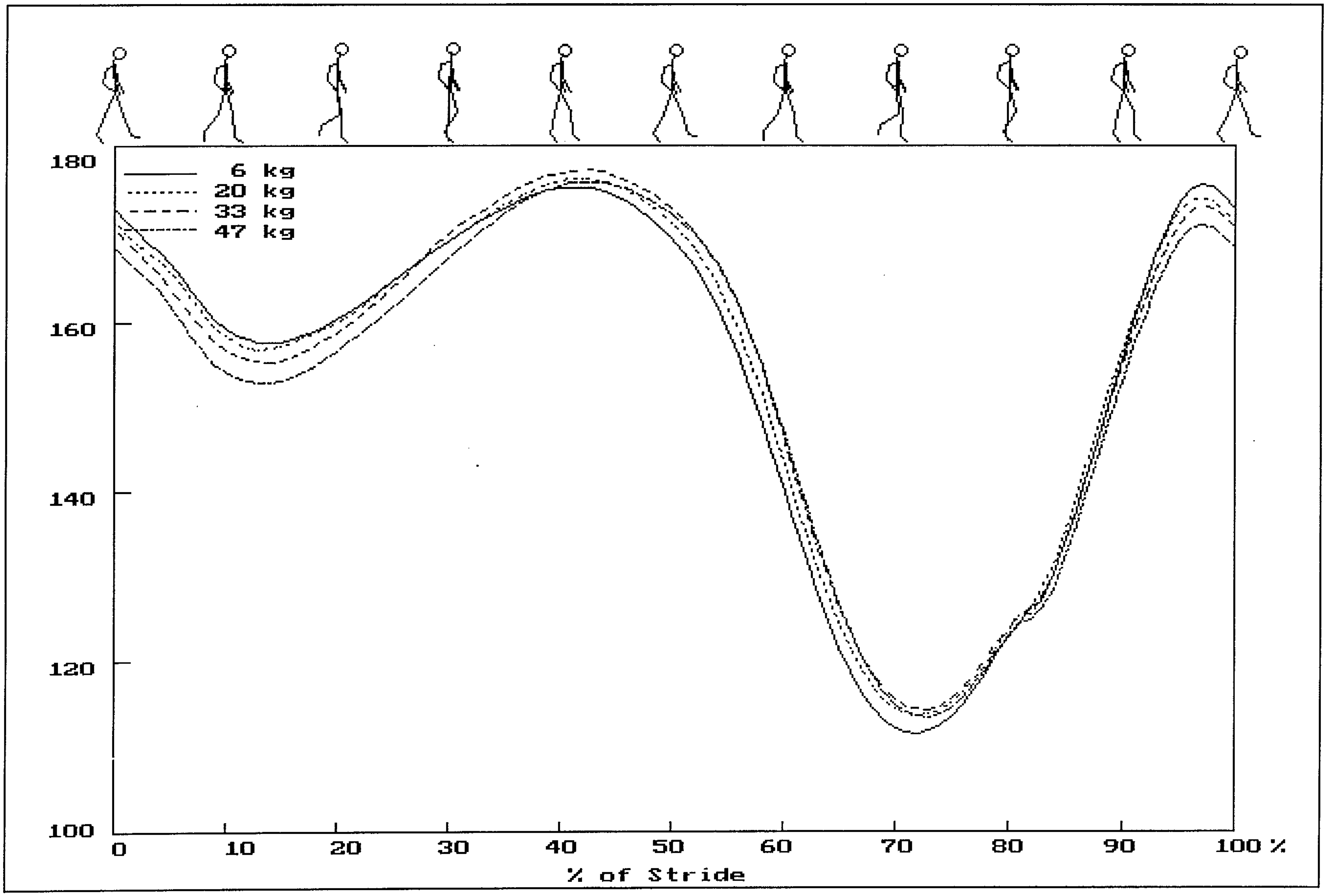


Figure 5. Load effects for knee angle (degrees).

Upper Body Sagittal Plane Ranges of Motion

There were significant load effects for all upper body ranges of motion except for minimum elbow angle, elbow range of motion, and maximum shoulder angle (Table 4). Maximum elbow angle at 6 kg was significantly greater than at all three heavier loads, which did not differ significantly among each other.

Minimum shoulder angle at 6 kg was significantly greater than at all three heavier loads, which did not differ significantly among each other. The only significant difference for shoulder range of motion was between the 6 and 47 kg loads, of which the latter was only 2/3 that of the former. Increase in load did not affect the forward swing of the arms. Thus, the difference in range of motion was attributable mainly to a decrease of 60% in degree of rearward arm swing as the load increased from 6 to 47 kg.

Both minimum and maximum trunk angle dropped significantly with each load increment. Trunk range of motion increased with load, and was significantly greater for 33 and 47 kg than for 6 kg. The difference translated to an increase of 22% in trunk sagittal sway from the lightest to heaviest load.

Table 4. Upper body sagittal plane ranges of motion at loads of 6, 20, 33, and 47 kg (mean±SD)

Variables	6 Kg	20 Kg	33 Kg	47 Kg	Prob.
Minimum elbow angle (rad)	2.52±0.18	2.53±0.10	2.49±0.11	2.48±0.11	0.065
Maximum elbow angle (rad)	2.87±0.10	2.84±0.09†	2.82±0.15†	2.75±0.12†	0.001*
Elbow range of motion (rad)	0.33±0.13	0.34±0.26	0.33±0.18	0.28±0.12	0.347
Minimum shoulder angle (rad)	-0.19±0.11	-0.12±0.10†	-0.11±0.09†	-0.08±0.13†	0.000*
Maximum shoulder angle (rad)	0.26±0.15	0.28±0.20	0.24±0.18	0.22±0.15	0.361
Shoulder range of motion (rad)	0.45±0.21	0.40±0.26	0.35±0.22	0.30±0.20†	0.005*
Minimum trunk angle (rad)	1.47±0.08	1.40±0.08†	1.35±0.08†‡	1.30±0.07†‡§	0.000*
Maximum trunk angle (rad)	1.54±0.07	1.47±0.08†	1.42±0.07†‡	1.38±0.07†‡§	0.000*
Trunk range of motion (rad)	0.065±0.02	0.070±0.02	0.077±0.02†	0.079±0.02†	0.001*

* statistically significant ($p<0.05$) load effect - ANOVA

† significantly ($p<0.05$) different from 6 kg - Tukey post-hoc test

‡ significantly ($p<0.05$) different from 20 kg - Tukey post-hoc test

§ significantly ($p<0.05$) different from 33 kg - Tukey post-hoc test

Center of Mass Parameters

There were no significant load effects for maximum center of mass horizontal, upward, and downward velocities (Table 5). However, minimum horizontal velocity increased with load, and was significantly greater at 47 kg than at 6 kg, with no other significant inter-load differences. Both minimum and maximum vertical position of the body center of mass (not including the pack) decreased significantly as load increased (Figure 6). For maximum vertical position, all inter-load differences were significant except between 33 and 47 kg. For minimum vertical position, all inter-load differences were significant except between 6 and 20 kg.

Table 5. Center of mass parameters at loads of 6, 20, 33, and 47 kg (mean±SD)

Variables	6 Kg	20 Kg	33 Kg	47 Kg	Prob.
Maximum horizontal velocity (m/sec)	1.65±0.20	1.67±0.20	1.66±0.18	1.69±0.17	0.450
Minimum horizontal velocity (m/sec)	0.86±0.14	0.90±0.15	0.90±0.14	$0.94\pm0.18^{\dagger}$	0.012^{*}
Maximum upward velocity (m/sec)	0.32±0.08	0.35±0.13	0.36±0.14	0.33±0.07	0.149
Maximum downward velocity (m/sec)	0.34±0.09	0.33±0.10	0.34±0.10	0.31±0.07	0.453
Maximum vertical position (m)	1.01±0.06	$1.00\pm0.06^{\dagger}$	$0.98\pm0.06^{\dagger\ddagger}$	$0.98\pm0.06^{\dagger\ddagger}$	0.000^{*}
Minimum vertical position (m)	0.95±0.06	0.95±0.07	$0.93\pm0.06^{\dagger\ddagger}$	$0.93\pm0.06^{\dagger\ddagger}$	0.000^{*}

* statistically significant ($p<0.05$) load effect - ANOVA
† significantly ($p<0.05$) different from 6 kg - Tukey post-hoc test
‡ significantly ($p<0.05$) different from 20 kg - Tukey post-hoc test

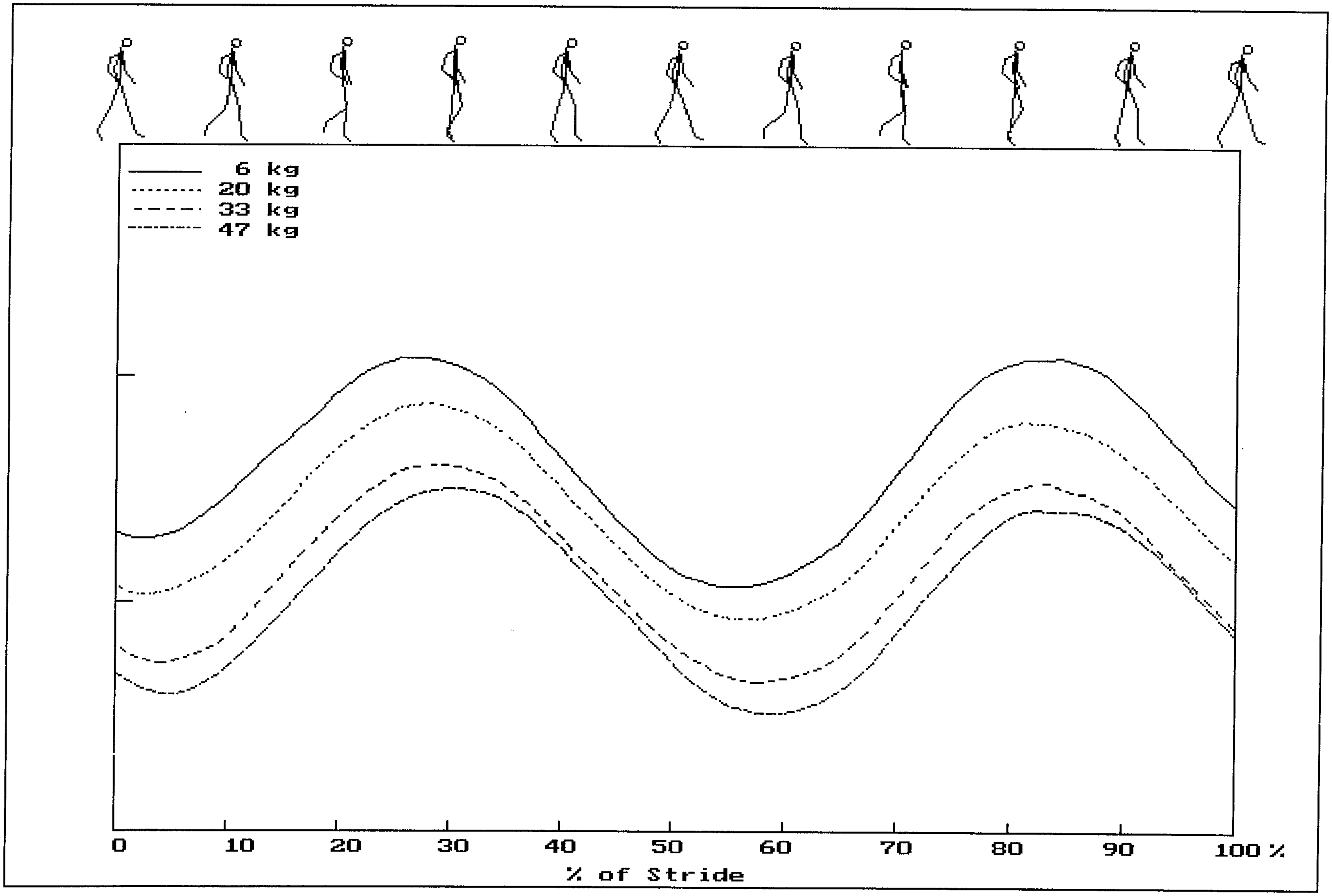


Figure 6. Load effects for the center of mass vertical position (cm).

Ground Reaction Forces and Impulses

Vertical, front-back, and medio-lateral ground reaction forces are respectively depicted in Figures 7, 8, and 9, all of which show generally greater forces with increasing load. There were significant load effects on the impulses and average forces along all three axes, except for lateral impulse and average force (Table 6). Propulsive, braking, and vertical impulses and average forces increased significantly with each increment in test load. Medial impulse and average medial force at the 33 and 47 kg loads were significantly greater than at the 6 kg load, with no other significant inter-load differences.

Table 6. Ground reaction forces at loads of 6, 20, 33, and 47 kg (mean±SD)

Variables	6 Kg	20 Kg	33 Kg	47 Kg	Prob.
Propulsive impulse (N•s)	29.1±4.99	33.4±5.37†	38.6±6.44†‡	40.9±6.79†‡§	0.000*
Braking impulse (N•s)	28.0±4.90	32.3±5.31†	38.3±5.70†‡	42.7±5.48†‡§	0.000*
Lateral impulse (N•s)	1.57±0.86	1.60±1.06	1.72±1.21	1.98±1.24	0.051
Medial impulse (N•s)	17.6±6.24	19.3±5.83	22.2±8.18†	22.1±9.33†	0.004*
Vertical impulse (N•s)	480±65	548±68†	629±73†‡	686±77†‡§	0.000*
Average propulsive force (N)	69.7±12.7	82.4±16.1†	92.9±14.6†‡	103.9±17.4†‡§	0.000*
Average braking force (N)	74.6±14.4	83.8±14.6†	98.1±18.1†‡	105.4±13.9†‡§	0.000*
Average lateral force (N)	12.7±4.88	12.3±5.36	12.3±6.10	13.4±5.08	0.510
Average medial force (N)	25.9±7.88	28.5±7.18	32.3±10.1†	33.0±11.8†	0.000*
Average vertical force (N)	598±62	685±62†	771±60†‡	852±71†‡§	0.000*

* statistically significant ($p<0.05$) load effect - ANOVA
† significantly ($p<0.05$) different from 6 kg - Tukey post-hoc test
‡ significantly ($p<0.05$) different from 20 kg - Tukey post-hoc test
§ significantly ($p<0.05$) different from 33 kg - Tukey post-hoc test

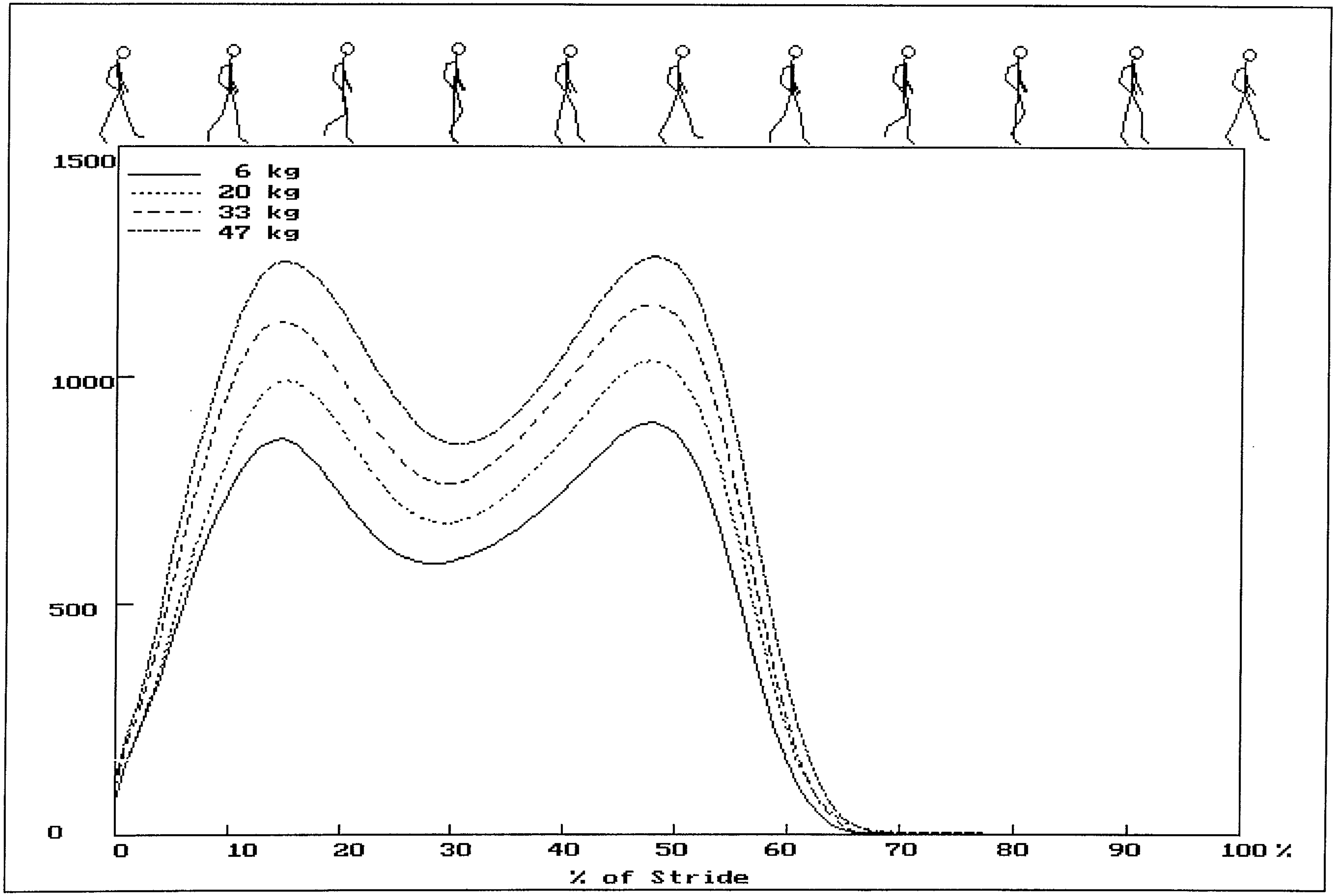


Figure 7. Load effects for vertical ground reaction force (N).

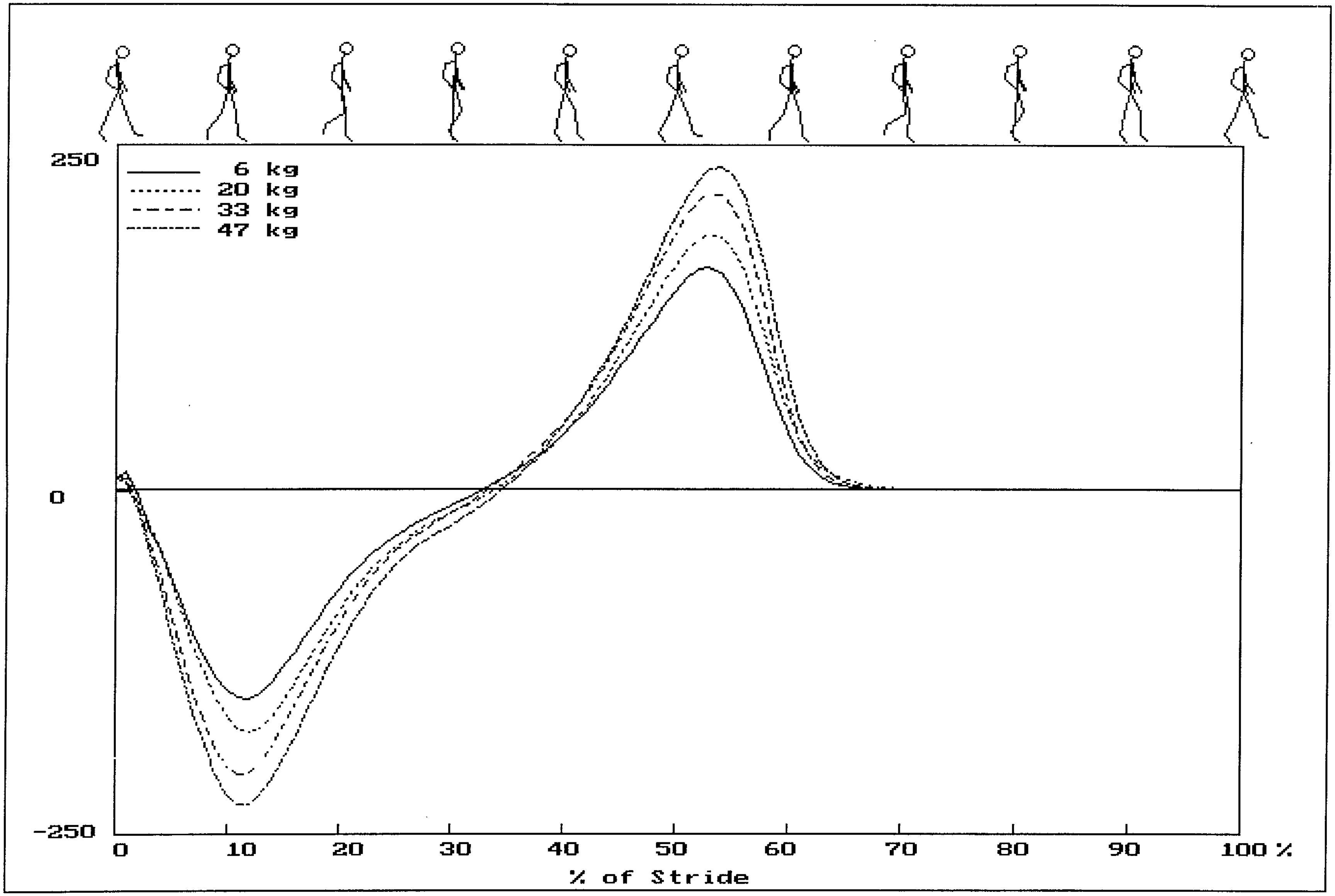


Figure 8. Load effects for front-back ground reaction force (N, positive = propulsive, negative = braking).

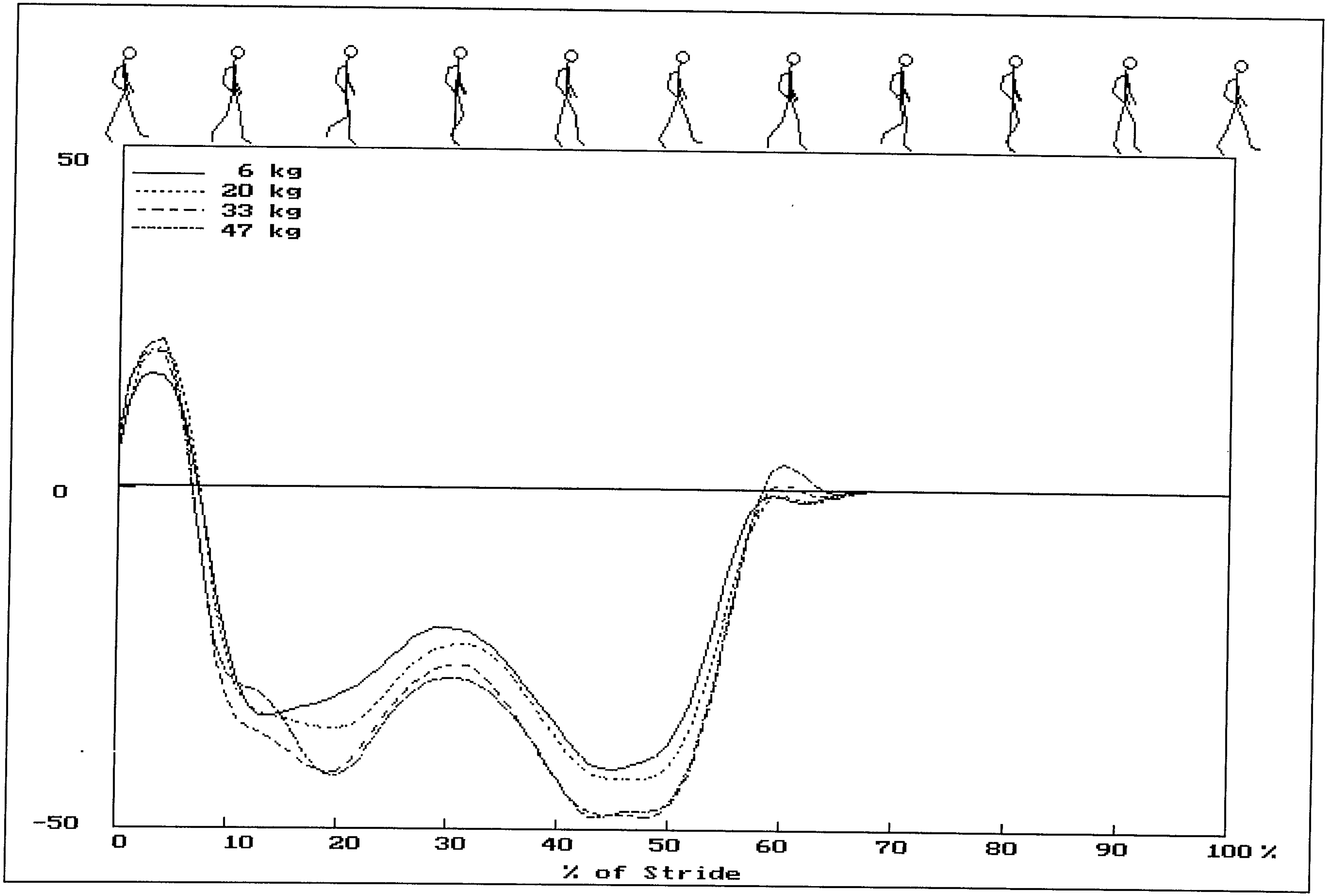


Figure 9. Load effects for medio-lateral ground reaction force (N).

There were significant load effects for all peak forces except in the lateral direction (Table 7). The front-back ground reaction force showed an initial small propulsive impulse at heel-strike followed by a much larger braking impulse followed by a second propulsive impulse, this one of similar magnitude to the braking impulse. Peak propulsive and braking forces, as well as the first and second peak vertical forces, increased significantly with each increment in test load. Peak medial force increased with load and was significantly greater for 33 and 47 kg than for 6 kg, with no other significant inter-load differences. A small lateral ground reaction impulse occurred at the beginning of the stance phase, and was not affected by the load. As to percentage of stride at which peak forces occurred, there were no significant load effects for peak braking, lateral, and medial forces. However, peak propulsive force at the 47 kg load occurred at a significantly greater percentage of stride than at the 6 and 20 kg loads, with no other significant inter-load differences. The vertical ground reaction force curve had two peaks of which the second peak was higher than the first (Figure 5). First peak vertical force at the 6 kg load occurred at a lesser percentage of stride than at the three heavier loads, with no other significant inter-load differences. The 2nd peak vertical force at the 47 kg load occurred at a larger percentage of the stride than at the 20 kg load, with no other significant inter-load differences.

Table 7. Peak ground reaction forces and timing at loads of 6, 20, 33, and 47 kg (mean±SD)

Variables	6 Kg	20 Kg	33 Kg	47 Kg	Prob.
Peak propulsive force (PPF, N)	168±29.0	196±34.3†	223±31.2†‡	243±36.2†‡§	0.000*
% of stride at PPF	54.1±1.73	54.3±1.50	54.7±1.68	55.2±1.57†‡	0.001*
Peak braking force (PBF, N)	158±32.9	180±31.4†	214±42.2†‡	237±32.9†‡§	0.000*
% of stride at PBF	12.4±1.18	12.4±1.17	12.3±1.11	12.2±1.46	0.752
Peak lateral force (PLF, N)	24.5±10.1	25.5±11.9	26.8±14.0	29.1±13.8	0.146
% of stride at PLF	4.27±1.77	4.10±1.46	4.25±1.55	4.29±1.71	0.914
Peak medial force (PMF, N)	46.8±12.5	50.0±11.6	55.3±17.4†	56.8±19.5†	0.001*
% of stride at PMF	39.0±13.7	36.3±14.9	38.4±13.7	36.4±141.	0.742
1st Peak vertical force (PVF1, N)	876±106	1,000±111†	1,137±118†‡	1,264±124†‡§	0.000*
% of stride at PVF1	15.1±1.32	15.9±1.77†	15.8±1.82†	16.1±1.62†	0.000*
2nd Peak vertical force (PVF2, N)	906±112	1040±120†	1,162±120†‡	1,270±143†‡§	0.000*
% of stride at PVF2	48.5±1.50	48.4±1.22	48.9±1.56	49.3±1.72‡	0.027*

* statistically significant ($p<0.05$) load effect - ANOVA
† significantly ($p<0.05$) different from 6 kg - Tukey post-hoc test
‡ significantly ($p<0.05$) different from 20 kg - Tukey post-hoc test
§ significantly ($p<0.05$) different from 33 kg - Tukey post-hoc test

Joint Torques and Forces

Figures 10, 11, and 12 respectively depict torque about the ankle, knee, and hip during a full stride. There were significant load effects for all joint torques except peak ankle dorsiflexion torque, and peak knee flexion torque, with all torques increasing with load (Table 8). Peak ankle plantarflexion torque increased significantly with each increment in load except from 20 to 33 kg. Peak knee extension torque increased significantly with each increment in load except from 6 to 20 kg. Peak hip extension torque increased with load, with significant differences only between 6 and 33 kg, 6 and 47 kg, and between 20 and 47 kg. Peak hip flexion torque increased with load, with a significant difference only between the 6 and 47 kg loads. Timing of torque application about the ankle and hip did not seem to be affected by load. However, at the knee, greater load appeared to be associated with later second peak for knee extension torque.

There were significant load effects for all joint forces except peak upward shank-on-foot force with all but peak upward trunk-on-hip force increasing with load (Table 8). The peak joint force variables that increased significantly with each increment in load included forward shank-on-foot force, backward shank-on-foot force, downward shank-on-foot force, forward thigh-on-shank force, backward thigh-on-shank force, downward thigh-on-shank force, forward trunk-on-hip force, backward trunk-on-hip force, and downward trunk-on-hip force. Peak upward trunk-on-hip force did not change in a consistent direction as load increased and was significantly different only between the 30 and 47 kg loads.

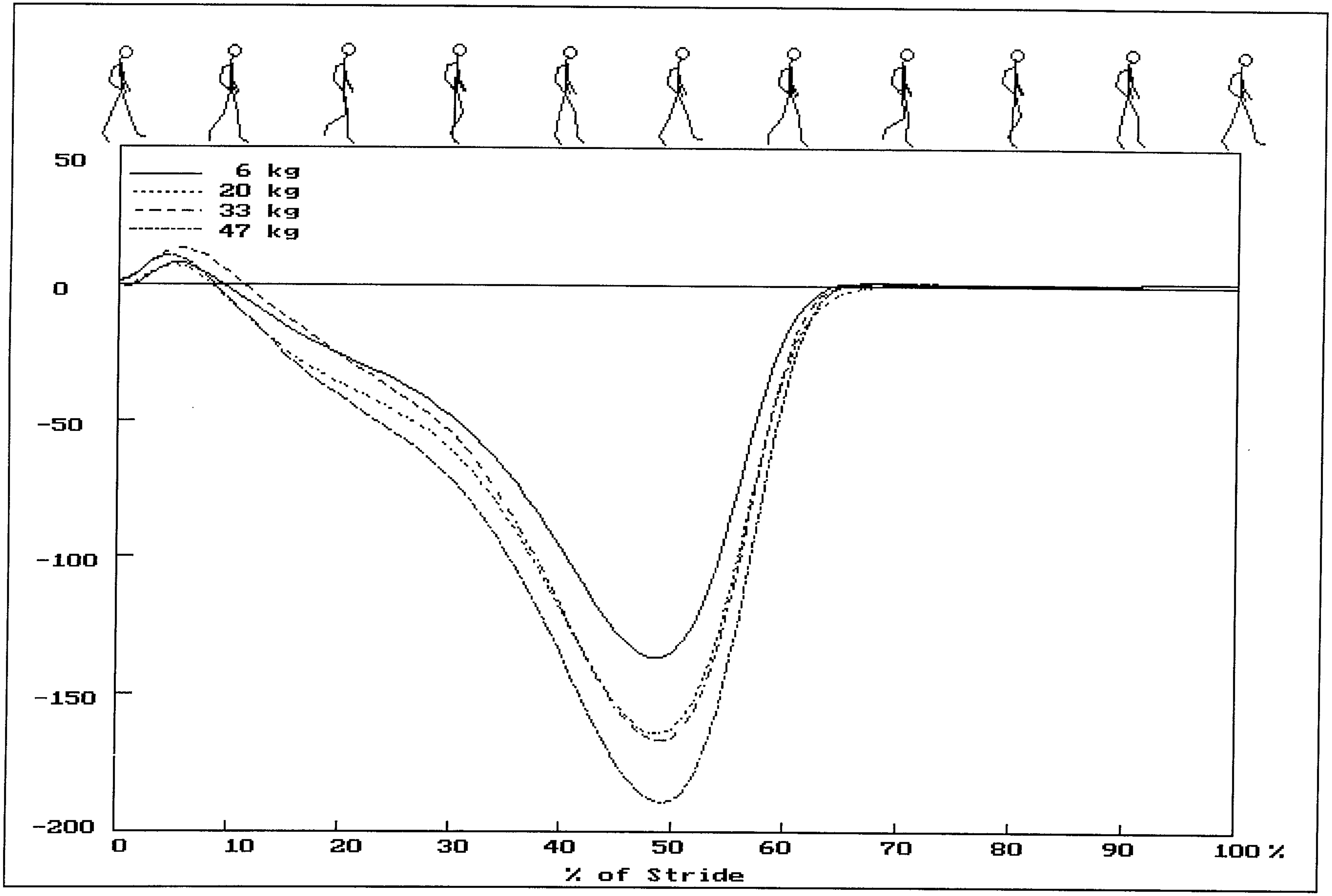


Figure 10. Load effects for ankle torque (N•m).

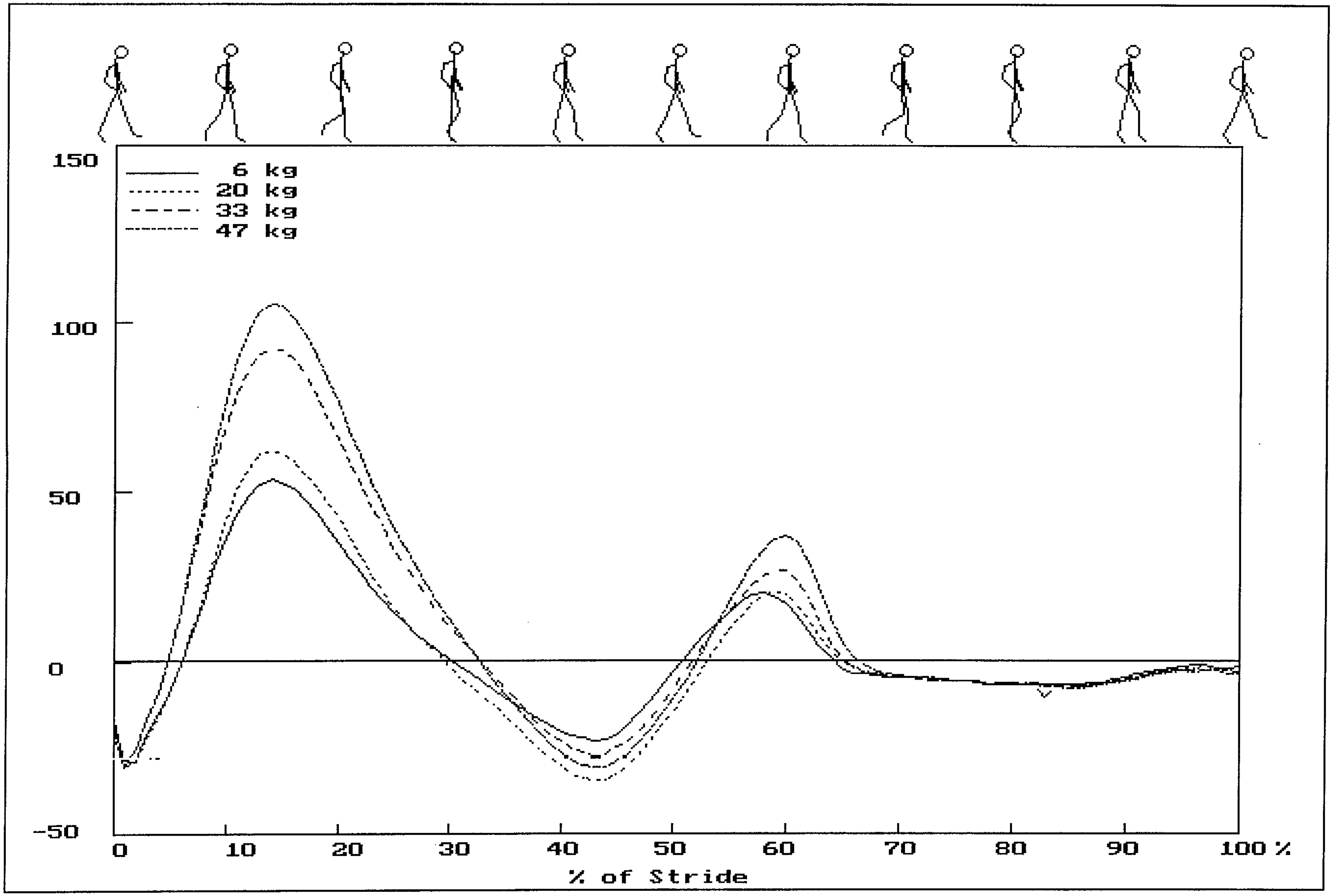


Figure 11. Load effects for knee torque (N•m).

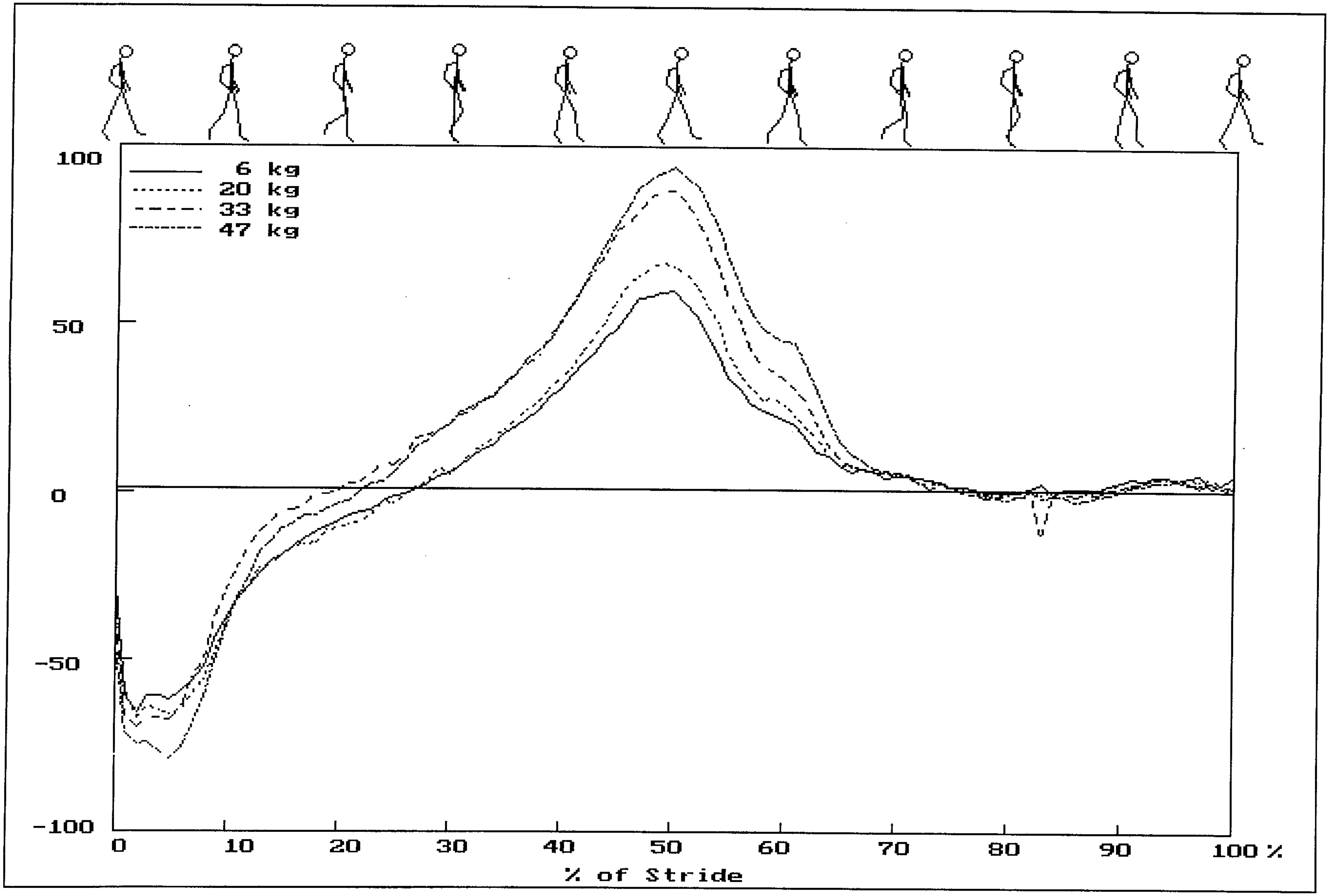


Figure 12. Load effects for hip torque (N•m).

Table 8. Joint forces and torques at loads of 6, 20, 33, and 47 kg (mean±SD)

Variables	6 Kg	20 Kg	33 Kg	47 Kg	Prob.
peak forward shank-on-foot force (N)	153±32.1	$176 \pm 30.9^{\dagger}$	$210 \pm 40.9^{\dagger\ddagger}$	$231 \pm 32.0^{\dagger\ddagger\S}$	0.000^{*}
peak backward shank-on-foot force (N)	160±27.4	$187 \pm 27.3^{\dagger}$	$214 \pm 29.9^{\dagger\ddagger}$	$233 \pm 34.5^{\dagger\ddagger\S}$	0.000^{*}
peak upward shank-on-foot force (N)	14.4±1.95	14.4±1.83	14.6±2.00	14.5±1.82	0.573
peak downward shank-on-foot force (N)	907±110	$1{,}040 \pm 118^{\dagger}$	$1{,}184 \pm 116^{\dagger\ddagger}$	$1{,}314 \pm 126^{\dagger\ddagger\S}$	0.000^{*}
peak ankle dorsiflexion torque (N•m)	9.18±5.72	8.35±6.34	9.02±5.52	9.53±6.1	0.979
peak ankle plantarflexion torque (N•m)	139±20.6	$167 \pm 31.0^{\dagger}$	$177 \pm 34.2^{\dagger}$	$192 \pm 33.5^{\dagger\ddagger\S}$	0.000^{*}
peak forward thigh-on-shank force (N)	141±30.1	$164 \pm 30.1^{\dagger}$	$199 \pm 38.4^{\dagger\ddagger}$	$218 \pm 30.0^{\dagger\ddagger\S}$	0.000^{*}
peak backward thigh-on-shank force (N)	138±24.4	$165 \pm 30.2^{\dagger}$	$192 \pm 27.0^{\dagger\ddagger}$	$209 \pm 31.4^{\dagger\ddagger\S}$	0.000^{*}
peak upward thigh-on-shank force (N)	56.5±7.94	55.8±7.37	56.4±7.01	54.9±6.63	0.113
peak downward thigh-on-shank force (N)	870±106	$1{,}003 \pm 114^{\dagger}$	$1{,}150 \pm 112^{\dagger\ddagger}$	$1{,}277 \pm 122^{\dagger\ddagger\S}$	0.000^{*}
peak knee extension torque (N•m)	54.3±19.5	66.8±19.2	$81.3 \pm 25.8^{\dagger\ddagger}$	$107.4 \pm 40.0^{\dagger\ddagger\S}$	0.000^{*}
peak knee flexion torque (N•m)	40.7±16.0	43.1±12.2	50.2±20.4	47.6±20.6	0.055
peak forward trunk-on-hip force (N)	131±29.8	$164 \pm 44.4^{\dagger}$	$192 \pm 36.8^{\dagger\ddagger}$	$211 \pm 32.4^{\dagger\ddagger\S}$	0.000^{*}
peak downward trunk-on-hip force (N)	128±26.6	$155 \pm 33.3^{\dagger}$	$179 \pm 24.7^{\dagger\ddagger}$	$194 \pm 30.1^{\dagger\ddagger\S}$	0.000^{*}
peak upward trunk-on-hip force (N)	144±22.7	142±20.1	152±32.6	$140 \pm 18.8^{\S}$	0.029^{*}
peak downward trunk-on-hip force (N)	837±106	$970 \pm 116^{\dagger}$	$1{,}126 \pm 118^{\dagger\ddagger}$	$1{,}243 \pm 124^{\dagger\ddagger\S}$	0.000^{*}
peak hip extension torque (N•m)	66.6±26.4	75.5±34.3	$87.8 \pm 44.3^{\dagger}$	$98.1 \pm 46.6^{\dagger\ddagger}$	0.000^{*}
peak hip flexion torque (N•m)	80.4±26.6	85.2±25.8	89.6±30.8	$94.3 \pm 24.8^{\dagger}$	0.021^{*}

* statistically significant ($p<0.05$) load effect - ANOVA

† significantly ($p<0.05$) different from 6 kg - Tukey post-hoc test

‡ significantly ($p<0.05$) different from 20 kg - Tukey post-hoc test

§ significantly ($p<0.05$) different from 33 kg - Tukey post-hoc test

Muscle Electrical Activity

As shown in Table 9, EMG amplitude averaged over a stride increased significantly with load for the trapezius, spinal erector, quadriceps, and gastrocnemius. For the trapezius, amplitude rose with each increment in load, and all inter-load differences were significant except between 20 and 33 kg. For the spinal erector, average EMG was lowest for the 20 kg load and highest at 47 kg, with significant differences only between 20 and 47 kg. For the quadriceps, EMG amplitude rose with each increment in load, with inter-load differences significant between 6 and 33 kg, 6 and 47 kg, and 20 and 47 kg. For the gastrocnemius, average EMG amplitude increased with load up to 33 kg, but did not increase further when the load increased to 47 kg; differences were significant between 6 and 33 kg, 6 and 47 kg, and 20 and 33 kg. As seen in Figure 13, EMG amplitude peaked with contra-lateral heel-strike.

Table 9. Muscle activity at loads of 6, 20, 33, and 47 kg (mean±SD)

Variables	6 Kg	20 Kg	33 Kg	47 Kg	Prob.
Average trapezius EMG (μV)	26.5±15.0	41.4±27.2†	52.2±29.1†	66.0±50.5†‡§	0.000*
Average spinal erector EMG (μV)	82.3±40.7	67.8±48.8	79.7±51.5	96.0±50.5‡	0.005*
Average quadriceps EMG (μV)	57.0±24.3	80.7±55.9	110.6±72.8†	128.7±83.8†‡	0.000*
Average hamstrings EMG (μV)	89.2±39.3	104.8±63.9	99.1±73.5	121.9±134.0	0.314
Average tibialis anterior EMG (μV)	263±106	272±140	279±135	294±126	0.656
Average gastrocnemius EMG (μV)	178±98	198±102	239±108†‡	235±129†	0.000*

Note - Table values are EMG amplitudes averaged over one full stride.
* statistically significant ($p<0.05$) load effect - ANOVA
† significantly ($p<0.05$) different from 6 kg - Tukey post-hoc test
‡ significantly ($p<0.05$) different from 20 kg - Tukey post-hoc test
§ significantly ($p<0.05$) different from 33 kg - Tukey post-hoc test

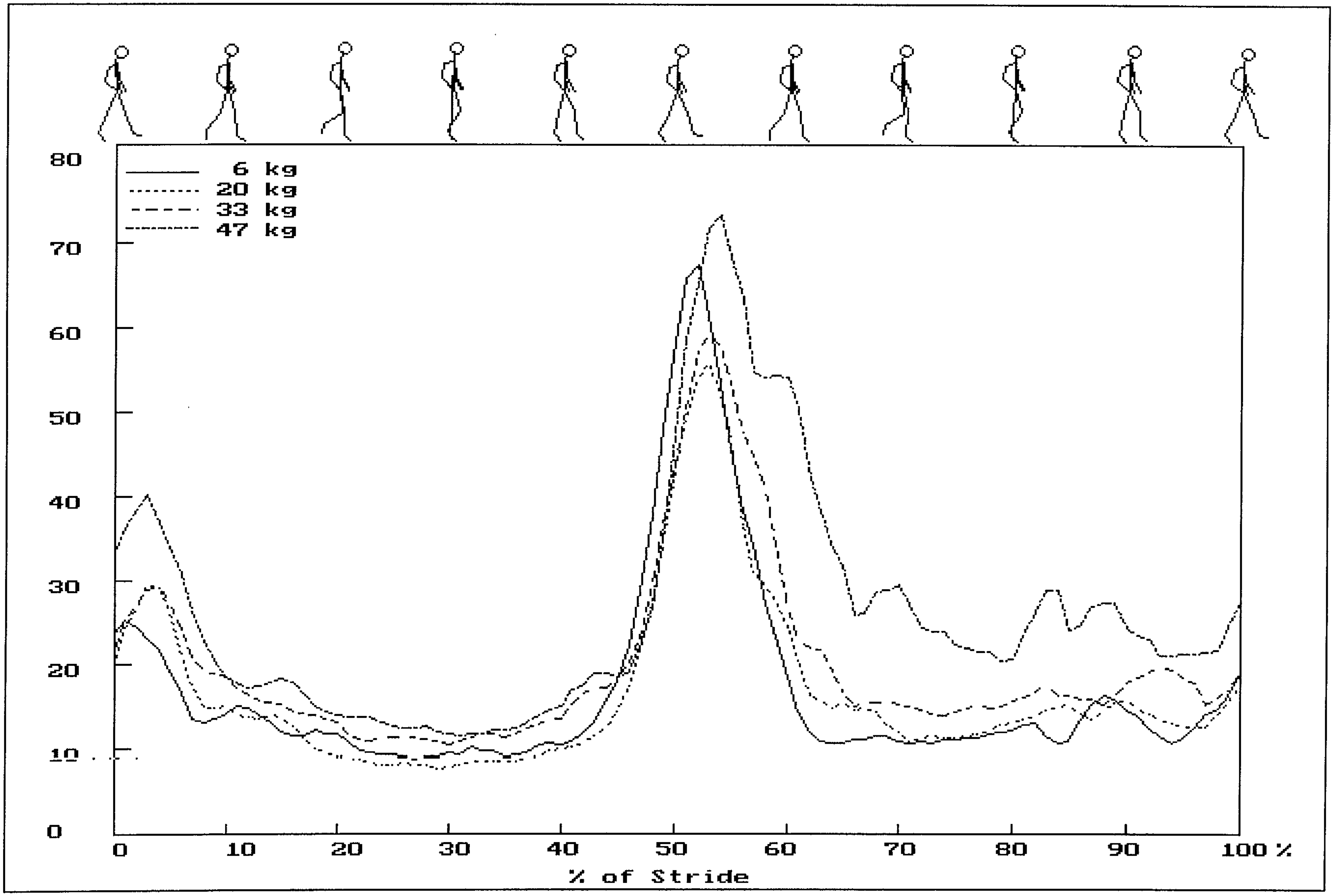


Figure 13. Load effects for spinal erector EMG (μV).

Backpack Acceleration

Load significantly affected peak backward and downward acceleration, but not peak forward and upward acceleration, as seen in Table 10. For peak downward acceleration, the significant inter-load differences were between 6 and 47 kg, and between 20 and 47 kg. Peak backward acceleration increased with each increment in load, with significant differences among all loads except between 6 and 20 kg.

Table 10. Backpack acceleration at loads of 6, 20, 33, and 47 kg (mean±SD)

Variables	6 Kg	20 Kg	33 Kg	47 Kg	Prob.
Peak upward acceleration (g)	0.41±0.12	0.41±0.10	0.41±0.11	0.46±0.18	0.137
Peak downward acceleration (g)	0.33±0.09	0.33±0.09	0.36±0.11	0.38±0.11†‡	0.006*
Peak backward acceleration (g)	0.48±0.12	0.51±0.14	0.61±0.19†‡	0.76±0.19†‡§	0.000*
Peak forward acceleration (g)	0.08±0.08	0.08±0.10	0.07±0.11	0.04±0.10	0.367

* statistically significant ($p<0.05$) load effect - ANOVA
† significantly ($p<0.05$) different from 6 kg - Tukey post-hoc test
‡ significantly ($p<0.05$) different from 20 kg - Tukey post-hoc test
§ significantly ($p<0.05$) different from 33 kg - Tukey post-hoc test

DISCUSSION

LOAD EFFECTS

Stride Parameters

The observed increase in stride frequency and concomitant decrease in stride time when the load was increased to 47 kg have some functional value. Kinoshita (44) observed that with increasing load, dorsiflexion of the foot at toe-off increased with load. He felt this could expose the metatarsal bones to greater and more prolonged mechanical stress, possibly accounting for the incidence of stress fractures of the foot among backpackers and soldiers. He hypothesized that step length should be shortened and stride frequency increased as load increases in order to avoid such stress on the bones of the foot. Not all previous studies have found stride frequency to increase with increasing load (44). Martin and Nelson (52) found that women consistently increased stride frequency with load while men did not. Our previous study of female soldiers (33) showed an increase in stride frequency as the load increased. The percentage increase was similar to that found in the present study. The actual stride frequency of those females was 10%-13% higher than that of the males whose data are reported herein. That is to be expected given the disparity in height between males and females.

The percentage of stride at which toe-off occurs can also be called the percentage of stride consisting of stance phase. By whichever name, this variable is directly related to double-support expressed as percentage of stride, assuming left-right symmetry of gait. Percentage of stride at toe-off minus 50% times two equals percentage of stride under double-support. Therefore, it is not surprising that both percentage of stride at toe-off and percentage of stride under double-support both changed in the same direction as load increased. The increase in double-support with load, especially for the heaviest load, has the obvious benefit of providing greater control and stability as load increases. Martin and Nelson (52) also found an increase in percentage of stride under double-support as the load increased. In one of our previous load carriage studies (33), female soldiers increased their percentage of stride in double-support as the load increased, and to a similar degree to that found in the present study. In another of our studies (59), men responded to the heaviest load by increasing the percentage of stride under double-support.

Lower Body Sagittal Plane Ranges of Motion

When the stride is considered to start at heel-strike, the sole of the foot hits the ground at about 6% of stride, at which time the greatest degree of plantarflexion of the mid-stance phase occurs (82). The subsequent apparent ankle dorsiflexion (Fig. 1) results from the continued forward movement of the lower leg while the foot remains flat against the ground, which is required to shift the body's center of mass horizontally forward towards the front foot.

The knee started flexing from an almost fully extended position just before heel-strike, and continued flexing until shortly after the sole of the foot hit the ground. As the load

increased, the knee operated over an increasingly flexed range of motion as it flexed during the shock absorption phase from heel-strike until mid-stance. The greater knee flexion helped keep the center of mass lower, thus increasing stability as the load increased. In contrast, just after mid-stance, the knee was increasingly straight over the first three loads, possibly serving to lessen the increase in quadriceps tension with increasing load. However, the trend was apparently overridden for the 47 kg load, where at mid-stance the knee was flexed as much as at the 20 kg load, possibly because of the difficulty of resisting the relatively great downward inertia.

The hip was at its greatest degree of flexion at heel-strike, after which it moved towards its maximum degree of extension that occurred at heel-strike of the opposite foot. The increase in degree of hip flexion (angle between thigh and trunk) with increasing load was largely accounted for by the increased forward inclination of the trunk.

Center of Mass Parameters and Upper Body Movement

The decrease in both minimum and maximum center of mass height with increasing loads was mainly due to the greater forward inclination of the trunk. To prevent falling backwards as load on the back increased, the trunk must have been brought forward to keep the body-plus-backpack center of mass over the feet. However, the adjustment was not complete in that the center of mass of the upper-body-plus-pack was further rearward than the center of mass of the trunk alone when no load was carried (8,25). Thus, increasing abdominal muscle activity was probably needed to keep the trunk from rotating rearward as load increased.

Ground Reaction Forces and Impulses

The first and second peak vertical forces exhibited by the soldiers are quite consistent with those we found in our previously reported study of male soldiers carrying backpacks (34). However, in that study, the load was distributed over the body in the form of a backpack as well as equipment on a fighting vest (designed to hold grenades, ammunition and other supplies), rifle and boots, while in the present study, virtually all the weight carried was in the backpack.

The fact that all the ground reaction forces and impulses except for the lateral ground reaction force and impulse increased proportionately for the three lightest loads but not for the heaviest load suggested a protective adjustment. The increased knee flexion throughout the gait cycle observed with the heaviest load may have helped avoid excessive vertical ground force. The almost fully extended position of the knee at the second peak of vertical ground reaction force may have contributed to the higher magnitude of the second than the first peak for the three lightest loads. However, the two peaks were almost the same for the 47 kg load.

The finding of greater magnitude of the second than the first peak was in agreement with the results of Schneider and Chao (66) but not with those of Kinoshita (43, 44) who observed that the first peak was higher than the second. The discrepancy with Kinoshita's

data, which was presumably collected in Japan, might be due to cultural differences in walking technique or population differences in body size and proportions.

The initial propulsive impulse seen at heel-strike results from flexion at the knee, rather than extension at the hip, as the heel strikes the ground. The observed hamstring activity before and during heel-strike relates to the deceleration of knee extension and the initiation of knee flexion. The small lateral ground reaction impulse which occurred at the beginning of the stance phase was apparently due to swinging of the leg towards the midline of the body just before heel-strike in order to keep the center of support under the center of mass, thus avoiding instability. The fact that the lateral ground reaction impulse did not increase with load is probably because only the leg, not the whole body, was swung toward the body midline before heel-strike. The impulse was needed only to laterally decelerate the leg after heel contact, not to decelerate the load and body.

The fact that medial ground reaction force increased as expected over the first three loads but not between the next-heaviest and heaviest loads indicates a gait adjustment that limited the medial travel of the center of mass for the heaviest load. By limiting force exerted laterally on the ground, the volunteer avoided medial ground reaction force which would move the center of mass medially, thereby creating instability as the center of mass moved horizontally away from the supporting foot. When the load is light the muscles can more easily handle the destabilizing torque due to a load not directly over the foot. However, for a very heavy load such destabilizing torque can become very large, necessitating a gait adjustment to limit such instability.

Joint Torques and Forces

Although torques about the ankle, knee, and hip clearly increased with load over the stride, the amount of increase about the different joints varied. As body-plus-pack mass increased by 49%, ankle peak plantarflexion torque increased by 38%, knee peak extension torque increased by 98%, and hip peak extension torque increased by 47%. Thus while the change in hip extension torque was proportional to the load increase, the change in knee extension torque was twice that expected from the change in load alone, while the change in ankle plantarflexion torque was somewhat less than that expected from the change in load. The finding indicates that the quadriceps muscles assume a disproportional share of the burden during heavy load carriage, while the soleus and gastrocnemius assume less of the burden than expected.

The later second peak for knee extension torque observed with the increase in load was apparently related to the increase in the proportion of the stride in double-support as load increased. In order to maintain stability under heavier load, the walker made a gait adjustment to keep both feet on the ground over a greater percentage of the stride. A way of doing that was to delay pushing off, thus accounting for the later second peak torque.

Muscle Electrical Activity

The increase in average trapezius muscle EMG activity was more than three times the percentage change in total load, despite the fact that a waist belt designed to place as much of the weight as possible on the hips rather than the shoulders was part of the pack. Volunteers were checked to make sure the waist belt was tightly cinched. However, the design of the pack was apparently not adequate for maintaining the proportion of the entire load on the hips as the load increased. Large increases in trapezius muscle activity as the load increased were evidence for increased tension at the shoulders in response to the increased load.

The decrease noted in average spinal erector EMG activity when the load increased from 6 to 20 kg was consistent with the finding of other researchers. Lower spinal erector EMG amplitude has been observed when light to moderate loads were carried than without loads (8, 18, 19). This phenomenon has been attributed to a postural adjustment whereby the center of mass of the upper-body-plus-backpack was further rearward than the center of mass of the trunk alone when no load was carried (8, 25). Despite the fact that the trunk inclined forward as the load increased to keep the body-plus-backpack center of mass over the feet, the adjustment did not bring the center of mass as far forward over the foot as it was without a load. Yet when the load became much heavier, the center of mass apparently did not move further rearward relative to the foot. Thus the increased load in combination with the forward inclination of the trunk increased torque around the lower back that must have been counterbalanced by torque generated by the spinal erector muscles.

The spinal erector EMG electrode was located only on the right side of the back. Figure 13 makes it apparent that the right spinal erectors were most active at left heel-strike. The greater right side spinal erector muscle activity at and following left heel-strike can be accounted for by the fact that the ground-reaction braking force exerts a torque on the body which tends to twist the trunk in such way that the shoulder opposite the heel-strike foot moves forward. The spinal erectors on the side opposite the heel-strike foot must contract to prevent excessive twisting of the trunk during the stride. Such work by the spinal erectors increases with the greater inertia of trunk-plus-pack associated with a heavier backpack.

For about 5% of the stride right after the heel-strike, concentric knee flexion was evidenced by the concurrence of knee flexion with knee flexion torque and hamstring EMG activity. Immediately following, for about 8-10% of the stride ending at toe-off of the opposite foot, eccentric quadriceps activity was evidenced by continued knee flexion accompanied by quadriceps muscle activity and net knee extensor moment, which brought knee flexion to a halt. A period of concentric knee extension followed until mid-stance as knee extension moment continued. Later, after toe-off, the muscles around the knee relaxed, evidenced by a period of low quadriceps electrical activity, as the lower leg and foot swung forward propelled by gravity and inertia. At about 10% of the stride before heel-strike, the hamstring muscles became active again to decelerate the forward swing of the leg and foot and initiate heel-strike.

At heel-strike, the tibialis anterior, whose function is to exert dorsiflexion torque, acted eccentrically to control plantarflexion, thereby preventing the foot from slapping down on the ground. This was evidenced quite clearly by the surge in tibialis anterior muscle electrical activity just before heel-strike that reached a peak just after heel-strike, with relative quiescence throughout the rest of the stride. The lack of increase of tibialis anterior muscle activity with increasing load indicates that the muscle controls only the movement of the foot, and is thus not affected by the mass of the body-plus-load.

The gastrocnemius was quite inactive except for a period of high activity corresponding to the push-off phase, which occurred between mid-stance and heel-strike of the opposite foot, and included the second peak for vertical ground reaction force. Because gastrocnemius EMG activity increased along with the load up to 33 kg, but showed no further increase when the load was increased to 47 kg, the burden of carrying a very heavy load seemed to fall less on the calf muscles than on the muscles around the knee and hip. This corroborated the findings from joint torque calculations.

Backpack Acceleration

Peak downward acceleration of the pack increased along with the load. With the forward lean of the trunk, the spinal erector muscles could not resist the inertia of the increased load, and allowed gravity to accelerate the upper body and pack downward to a greater degree. The forward/downward velocity of the trunk in combination with a surge in erector spinae muscle activity when the trunk approached maximum forward lean indicated an eccentric phase wherein the back muscles controlled the rate of forward/downward acceleration of the trunk. The increased back muscle activity evidenced as the load increased was not enough to resist increased downward momentum of the pack. The increased trunk range of motion with the heavier load indicates that the trunk was allowed to move further forward before being brought to a stop.

Peak upward backpack acceleration was constant over the first three loads, but showed an average 12% increase with the heaviest load, which was not statistically significant. One might expect that the significant increase in downward acceleration with increased load would result in a compensatory high upward acceleration as load increased. The reason this apparently did not occur to a significant degree was probably that the upward impulse exerted by the body on the pack, needed to bring downward pack movement to a halt and begin upward pack movement, was not manifested via high peak forces, but rather by lower but more sustained forces.

The lack of load effect on peak forward backpack acceleration was probably due to the fact that much of the force accelerating the pack forward was transmitted from the load carrier to the pack through the shoulder straps. Due to the configuration of the straps, the trunk could move forward somewhat before the straps became taut enough to begin accelerating the pack to a significant degree. The result was that forward force on the pack was damped by the straps, causing the curve of forward force application on the pack to be flatter than would be expected from the forward acceleration of the trunk, thus diminishing

peak forward pack acceleration. This damping effect likely accounted for the lack of difference in peak forward pack acceleration with increasing load. Yet because the pack was already against the back as the trunk began to move backwards, there was little or no damping of backward forces exerted by the load carrier's back against the pack. Thus there was a significant load effect on peak backward backpack acceleration, which was 58% greater for the heaviest than the lightest load.

CONCLUSIONS

Based on the data analysis, the following conclusions were drawn:

1. The observed increase in stride frequency and decrease in stride time when the load became very heavy probably helped reduce the mechanical stress to the metatarsal bones. Such a gait adjustment may help to prevent stress fractures.

2. The observed increase with increasing load of double-support as percentage of stride, especially when going from the next-heaviest to the heaviest load, had the obvious benefit of providing greater control and stability. Delayed foot push-off with increasing load contributes to the increase in double-support as percentage of stride.

3. The greater knee range of motion with increasing load during the eccentric knee flexion period from heel-strike until mid-stance likely helped reduce shock to the body at and shortly after heel-strike.

4. The lower total body center of mass position as the load increased, effected both by greater knee flexion and more forward leaning trunk, helped control the potentially destabilizing effect of the load, by reducing the moment of inertia about the feet of the body-plus-load.

5. The initial propulsive impulse seen at heel-strike resulted from flexion at the knee, rather than extension at the hip, as the heel struck the ground, and was effected by hamstring muscle activity.

6. The lack of increase in medial ground reaction force between the next-heaviest and heaviest loads indicates a protective gait adjustment limiting the medial travel of the center of mass for very heavy loads. The resulting reduction in left-right body motion could help limit forces experienced by the relatively small muscles that adduct and control the rate of abduction of the hip.

7. As the load increases, hip extensor contribution increases proportionately. However, knee extensors' contribution increases more than expected, while ankle plantar flexor contribution increases somewhat less than expected.

8. The pack design used in the experiment, which incorporated a waist belt and frame, did not prevent the shoulders from supporting a considerable portion of the load by shifting it to the hips. As the load increased there was enough increased tension at the shoulders to require large increases in trapezius muscle activity.

9. Even though the trunk inclined forward as load increased to keep the center of body-plus-backpack mass over the feet, the adjustment did not bring the center of mass as far forward over the foot as without a load.

10. The spinal erectors produced their largest burst of activity at contralateral heel-strike, accounted for by the twisting of the trunk during the stride. A smaller burst at ipsilateral heel-strike was related to deceleration of the forward motion of the trunk as the body is braked at heel-strike, and the subsequent raising of the trunk.

11. The load carriage stride, in terms of the muscles that flexed and extended the knee, was characterized by concentric knee flexion from shortly before to shortly after heel-strike, eccentric knee flexion during a shock absorption phase, concentric knee extension during push-off, and a quiescent period after toe-off during the swing phase.

12. The load carriage stride, in terms of the muscles that dorsiflexed and plantarflexed the foot, was characterized by eccentric tibialis anterior activity at heel-strike which controlled the rate of plantarflexion to prevent the foot from slapping against the ground, which did not increase with load. The gastrocnemius was largely inactive except for a period of high activity during push-off, which occurred between mid-stance and heel-strike of the contralateral foot and included the second peak for vertical ground reaction force. Very heavy loads did not require additional contribution from the gastrocnemius; the burden of carrying a very heavy load seemed to fall less on the calf muscles than on the muscles around the knee and hip.

13. Trunk forward/downward excursion and acceleration increased with load. The erector spinae acted eccentrically to decelerate trunk motion as the trunk approached its maximum forward lean.

14. Load did not affect peak forward backpack acceleration because of the damping effect of the straps. Stretching of the straps enabled peak forward acceleration of the pack to occur later and be of lower magnitude than the peak forward acceleration of the trunk. However, a similar effect did not occur in regard to peak backward backpack acceleration of the trunk. As the trunk accelerated backwards, it pressed directly against the pack, with only slight damping due to compression of the soft tissue on the surface of the back. Therefore, load did not affect peak backward backpack acceleration to a significant degree.

RECOMMENDATIONS

LOAD CARRIAGE TECHNIQUE

1. An individual carrying a backpack load should employ a shorter stride length than during unloaded walking, and a concomitantly greater stride frequency. The benefits may include improved stability, lower joint torques during portions of the stride, and reduced mechanical stress to the metatarsal bones, which may help to prevent stress fractures. It must be noted, however, that reduced stride length and increased stride frequency may raise the energy cost of load carriage, particularly at higher walking speeds.

2. The percentage of stride under double-support should be increased along with the load in order to improve stability and spread the forces supporting the load over two legs rather than one for as long as possible. Double-support percentage should be increased by decreasing the swing phase as percentage of stride while maintaining the stance phase.

3. When a load is carried, it is desirable to keep the knees more bent throughout the stride than during unloaded walking in order to reduce shock to the body at heel-strike, lessen the distance that the load must be raised, and improve stability. As the load increases, the knees should be allowed to flex through a greater range of motion after heel-strike in order to improve shock absorption.

4. When carrying a heavy load, it is advisable to reduce the amount of lateral body motion that occurs with each step, in order to limit forces experienced by the relatively small muscles that adduct and control the rate of abduction of the hip.

PHYSICAL EXERCISES TO IMPROVE LOAD CARRIAGE PERFORMANCE

1. Because the knee extensor muscles contribute an increasing share of the burden as the load becomes heavier, strengthening of the quadriceps muscles through resistance exercise may help improve load carriage performance. Various forms of the squat exercise with a barbell provide an excellent strength-training stimulus for the quadriceps muscle groups. Other resistance exercises involving extension about the knees would also likely be beneficial. These include such weight-resisted exercises as the leg press, seated knee extension, Smith machine squat, and step-up onto a box or bench.

2. Both the spinal erector and abdominal muscles should be strengthened to improve load carriage ability. General exercises for the spinal erectors include the face-down trunk raise while holding a weight plate against the chest with the legs stabilized in either a horizontal or 45° position, the good-morning barbell trunk raise, and the barbell dead-lift. General exercises for the abdominals include sit-ups, curl-ups, and leg-raises (the first two can be weight-resisted). More specific supplemental exercises can be added which closely match the body segment ranges of motion

seen in load carriage. These would include relatively short range of motion (~10°) hip and trunk flexion and extension centered on a trunk position of about 7° of forward trunk inclination for a light load to about 14° of forward trunk inclination for a very heavy load. Limited range of motion sit-ups and face-down trunk raises with or without added weight would be beneficial. The trunk twisting observed during the stride may be simulated while performing such exercises.

3. Because the normal stride involves both concentric and eccentric muscle activity, exercise machines that have no eccentric phase are not recommended to improve load carriage performance. Such machines include those in which resistance is provided by air or fluid cylinders, and isokinetic or near-isokinetic machines that only provide concentric resistance.

4. The upper body plays a role in supporting the backpack. In particular, the trapezius muscles prevent depression of the shoulders while the pectoral muscles prevent the shoulders from being pulled back. The standing shoulder shrug exercise with a barbell or dumbbells is effective for developing the trapezius muscles. The bench press exercise is effective for developing the pectorals. To make the exercise particularly relevant to load carriage, it is advisable to thrust the shoulders ventrally after the weight is raised above the head, thereby providing a greater range of shoulder motion in the direction in which the shoulders must be pulled during backpacking.

LOAD CARRIAGE EQUIPMENT DESIGN

1. No matter what combination of military equipment (e.g., backpack, load carrying vest, body armor, weapons, ammunition, survival gear, communications devices), the soldier should be able to adjust the pack so that it comfortably distributes the load between the shoulders and the hip region, without excessive point pressures anywhere.

2. It may be beneficial to provide additional damping for deceleration of the pack's forward velocity. This may be accomplished by placing energy absorptive material between the pack and the back.

3. Because the second peak vertical ground reaction force (at push-off) is higher than the first (shortly after heel-strike), the addition of cushioning material in the shoe forefoot may help reduce possibly injurious forces to the skeleton. The cushioning material should be elastic so that absorbed energy can be utilized in propulsion, thereby avoiding an increase in the energy cost of load carriage.

RECOMMENDATIONS FOR FUTURE STUDY

Because few studies have been conducted on the biomechanics of load carriage, there are many areas that warrant investigation. Further research should be conducted on the effects on load carriage biomechanics of the following variables:

- Equipment
  - pack type
  - footwear type
  - clothing
  - load distribution
  - objects carried
- Individual differences
  - gender
  - height
  - body mass
  - body proportions
  - age
  - physical fitness
- Physical status
  - level of fatigue
  - female hormone cycle
  - hydration status
  - body temperature
  - mental state
- Technique
  - running vs. walking
  - stride length
  - stride frequency
- Environment
  - terrain type
  - surface incline
  - ambient temperature
  - ambient humidity
  - wind speed and direction
  - altitude

REFERENCES

1. Advanced Mechanical Technology, Inc. Model LG6-1-1 biomechanics platform instruction manual. Newton, MA, 1985.

2. Alexander, R. and A.S. Jays. Fourier analysis of forces exerted in walking and running. J Biomech, 13:383-390, 1980.

3. Balogun, J.A. Ergonomic comparison of three modes of load carriage. Int Arch Occ Environ Health, 58:35-46, 1986.

4. Balogun, J.A., R.J. Robertson, F.L. Goss, M.A. Edwards, R.C. Cox, and K.F. Metz. Metabolic and perceptual responses while carrying external loads on the head and by yoke. Ergonomics, 29:1623-1635, 1986.

5. Basmajian, J.V. The human bicycle. In: Biomechanics V-A, P. Komi (Ed.), University Park Press, Baltimore, MD, 1976, pp. 334-339.

6. Basmajian, J.V., and C.J. De Luca. Muscles Alive (5th Edition), Williams & Wilkins, Baltimore, MD, 1985.

7. Bloom, D. and A.P. Woodhull-McNeal. Postural adjustments while standing with two types of loaded backpack. Ergonomics, 30:1425-1430, 1987.

8. Bobet, J. and R.W. Norman. Effects of load placement on back muscle activity in load carriage. Eur J Appl Physiol, 53:71-75, 1984.

9. Boccardi, S., A. Pedotti, R. Ronado, and G.C. Santambrogio. Evaluation of muscular moments at the lower limb joints by an on-line processing of kinematic data and ground reaction. J Biomech, 14:35-45, 1981.

10. Borghols, E.M.A. Influence of heavy weight carried on the cardiorespiratory system during exercise. Eur J Appl Physiol, 38:161-169, 1978.

11. Brandell, B. Electromyographic-cinematographic study of the calf and quadriceps muscles during walking. In: Biomechanics V-A, P. Komi (Ed.), Baltimore, MD: University Park Press, 1976, pp. 319-327.

12. Brandell, B. Functional roles of the calf and vastus muscles in locomotion. Am J Physical Med, 56:59-74, 1977.

13. Bresler, B. and J.P. Frankel. The forces and moments in the leg during level walking. Transactions of the ASME, January:27-36, 1950.

14. Carter, R.C., R.S. Kennedy, and A.C. Bittner. Grammatical reasoning: a stable performance yardstick. Hum Factors, 23:587-591, 1981.

15. Catlin, M.J. and R.H. Dressendorfer. Effect of shoe weight on the energy cost of running. Med Sci Sports Exerc, 11:80, 1979.

16. Cavagna, G.A. and R. Margaria. Mechanics of walking. J Appl Physiol, 21:271-278, 1966.

17. Charteris, J, P.A. Scott, and J.W. Nottrodt. Metabolic and kinematic responses of African women headload carriers under controlled conditions of load and speed. Ergonomics, 32:1539-1550, 1989.

18. Cook, T.M and D.A. Neumann. An electromyographics analysis of the low back muscles during load carrying. Proceedings of the 1984 International Conference on Occupational Ergonomics, 556-558, 1984.

19. Cook, T.M. and D.A. Neumann. The effect of load placement on the EMG activity of the low back during load carrying by men and women. Ergonomics, 30:1413-1423, 1987.

20. Cotes, J.E. and F. Meade. Energy expenditure and mechanical energy demand in walking. Ergonomics, 3:97-119, 1960.

21. Datta, S.R. and N.L. Ramanathan. Ergonomic comparison of seven modes of carrying loads on the horizontal plane. Ergonomics, 14:269-278, 1971.

22. Dietz, V., D. Schmietbleicher, and J. Noth. Neuronal Mechanism of human locomotion. J Neurophysiol, 42:1212-1222, 1979.

23. Evans, O.M., Y. Zerbib, M.H. Faria, and H. Mono. Physiological responses to load holding and load carriage. Ergonomics, 26:161-171, 1983.

24. Falls, H.B. and L.D. Humphrey. Energy cost of running and walking in young women. Med Sci Sports, 8:9, 1976.

25. Floyd, W.F. and P. Silver. Function of the erectors spinae in flexion of the trunk. Lancet, 1:133-134, 1951.

26. Frykman, P.N., E.A. Harman, J.J. Knapik, and K.H. Han. Backpack vs. front-back pack: differential effects of fatigue on loaded walking posture. Med Sci Sports Exerc, 26(5):S140, 1994.

27. Fujita, M., N. Matsusaka, T. Norimatsu, R. Suzuki, and T. Chiba. The role of the ankle plantarflexors in level walking. In: Biomechanics IX-A, D.A. Winter, R.W. Norman, R.P. Wells, K.C. Hayes, and A.E. Patla (Eds.). Human Kinetics Publishers, Champaign, IL, 1983, pp. 484-488.

28. Gersten, J.W., W. Orr, A.W. Sexton, and D. Okin. External work in level walking. J Appl Physiol, 26:286-289, 1969.

29. Ghori, G.M.U. and R.G. Luckwill. Responses of the lower limb to load carrying in walking man. Eur J Appl Physiol, 54:145-150, 1985.

30. Goldman, R.F. and P.F. Iampietro, P.F. Energy cost of load carriage. J Appl Physiol, 17:675-676, 1962.

31. Gordon, M.J., B.R. Goslin, T. Graham, and J. Hoare. Comparison between load carriage and grade walking on a treadmill. Ergonomics, 26:289-298, 1983.

32. Grundy, M., P.A. Blackburn, R. Tosh, D. McLeish, and L. Smidt, L. An investigation of the centers of pressure under the foot while walking. J Bone Joint Surg, 57B:98-103, 1975.

33. Harman, E., P. Frykman, C. Pandorf, W. Tharion, R. Mello, J. Obusek, and J. Kirk. Physiological, biomechanical, and maximal performance comparisons of female soldiers carrying loads using prototype U.S. Marine Corps Modular Lightweight Load-Carrying Equipment (MOLLE) with Interceptor body armor and U.S. Army All-Purpose Lightweight Individual Carrying Equipment (ALICE) with PASGT body armor. USARIEM Technical Report T99-9, June 1999.

34. Harman, E., P. Frykman, C. Pandorf, W. Tharion, R. Mello, J. Obusek, and J. Kirk. Physiological, biomechanical, and maximal performance comparisons of soldiers carrying loads using U.S. Marine Corps Modular Lightweight Load-Carrying Equipment (MOLLE), and U.S. Army Modular Load System (MLS) prototypes. USARIEM Technical Report T99-4, February 1999.

35. Harman, E.A., J.P. Obusek, P.N. Frykman, C.J. Palmer, R.K Bills, and J. Kirk. Backpacking energy-cost and physical performance: internal vs. external frame, belt vs. no-belt. Med Sci Sports Exerc, 29(5):S205, 1997.

36. Harman, E.A., P.N. Frykman, J.J. Knapik, and K.H. Han. Backpack vs. front-back pack: differential effects of load on walking posture. Med Sci Sports Exerc, 26(5):S140, 1994.

37. Harman, E.A., K.H. Han, P. Frykman, M. Johnson, F. Russell, and M. Rosenstein. The effects on gait timing, kinetics, and muscle activity of various loads carried on the back. Med Sci Sports Exerc, 24:S129, 1992.

38. Hof, A.L., C.N.A. Pronk, and J.A. Van Bost. Comparison between EMG to force processing and kinetic analysis for the calf muscle moment in walking and stepping. J Biomech, 20:167-178, 1987.

39. Holewijn, M. Physiological strain due to load carrying. Eur J Appl Physiol, 61:237-245, 1990.

40. Inman, V.T. Human locomotion. J Can Med Assoc, 94:1047-1054, 1966.

41. Jacobs, N.A., J. Skorecki, and J. Charnley, J. Analysis of vertical component of force in normal and pathological gait. J Biomech, 5:11-34, 1972.

42. Jones, B.H., M.M. Toner, W.L. Daniels, and J.J. Knapik. The energy cost and heart-rate response of trained and untrained volunteers walking and running in shoes and boots. Ergonomics, 27:895-902, 1984.

43. Kinoshita, H. and B.T. Bates. Effects of two different load carrying system on ground reaction forces during walking. In: Biomechanics VIII-B, H. Matsui and K. Kobayashi (Eds.). Human Kinetics Publishers, Champaign, IL, 1981, pp. 574-581.

44. Kinoshita, H. Effects of different load and carrying systems on selected biomechanical parameters describing gait. Ergonomics, 28:1347-1362, 1985.

45. Knutsson, E. and C. Richards. Different types of disturbed motor control in gait and hemiparetic patients. Brain, 102:405-430, 1979.

46. Legg, S.J. and A. Mahanty. Energy cost of backpacking in heavy boots. Ergonomics, 29:433-438, 1986.

47. Luhtanen, P. and P. Komi. Force-, power-, and elasticity-velocity relationships in walking, running, and jumping. Eur J Appl Physiol, 44:279-289, 1980.

48. MacConaill, M.A. and J.V. Basmajian. Muscles and Movements Huntington, New York, Robert E. Krieger Publishing Company, 1977.

49. Mann, R.A. and J.L. Hagy. The initiation of gait. J Bone Joint Surg, 61A:232-239, 1979.

50. Mann, R.V. A kinetic analysis of sprinting. Med Sci Sports Exerc, 13:325-328, 1981.

51. Martin, P.E. Mechanical and physiological responses to lower extremity loading during running. Med Sci Sports Exerc, 17:427-433, 1985.

52. Martin, P.E. and R.C. Nelson. The effect of carried loads on the walking patterns of men and women. Ergonomics, 29:1191-1202, 1986.

53. Miller, J.F. and B.A. Stamford. Intensity and energy cost of weighted walking vs. running for men and women. J Appl Physiol, 62:1497-1501, 1987.

54. Munro, C.F., D.J. Miller, and A.J. Fuglevand. Ground reaction forces in running: a reexamination. J Biomech, 20:147-155, 1987.

55. Murray, M.P., A.B. Drought, and R.C. Kory. Walking patterns of normal men. J Bone Joint Surg, 46A:335-360, 1964.

56. Norman, R.W., D.A. Winter, and M.R. Pierrynowski. The utility of combining EMG and mechanical work rate data in load carriage studies. Proceedings of the 4th Congress of the International Society of Electrophysiological Kinesiology, 4:148-149, 1979.

57. Obusek, J.P., E.A. Harman, P.N. Frykman, C.J. Palmer, and R.K. Bills. The relationship of backpack center of mass location to the metabolic cost of load carriage. Med Sci Sports Exerc, 29(5):S205, 1997.

58. Pandolf, K.B., B. Givoni, and R.F. Goldman. Predicting energy expenditure with loads while standing or walking very slowly. J Appl Physiol, 43:577-581, 1977.

59. Pandorf, C.E., E.A. Harman, P.N. Frykman, J.P. Obusek, and T.J. Smith. A comparison of load carriage biomechanics of male and female soldiers. Med Sci Sports Exerc, 31(5):S190, 1999.

60. Pascoe, D.D. Influence of carrying book bags on gait cycle and posture of youths. Ergonomics, 40: 631-641, 1997.

61. Pedotti, A. A study of motor coordination and neuromuscular activities in human locomotion. Biological Cybernetics, 26:53-62, 1977.

62. Pierrynowski, M.R., R.W. Norman, and D.A. Winter. Mechanical energy analyses of the human during load carriage on a treadmill. Ergonomics, 24:1-14, 1981.

63. Ralston, H.J. and L. Lukin. Energy levels of human body segments during level walking. Ergonomics, 12:39-46, 1969.

64. Rosenrot, P., J.C. Wall, and J. Charteris. The relationship between velocity, stride time, support time, and swing time during normal walking. J Hum Movement, 6:323-335, 1980.

65. Samanta, A. and B.B. Chatterjee. Energy expenditure in manual load carriage. Industrial Health, 19:145-154, 1981.

66. Schneider, E. and E.Y. Chao. Fourier analysis of ground reaction forces in normals and patients with knee joint disease. J Biomech, 16:591-601, 1983.

67. Seroussi, R.E. and M.H. Pope. The relationship between trunk muscle electromyography and lifting moments in the sagittal and frontal planes. J Biomech 20:135-146, 1987.

68. Shiavi, R. and P. Griffin. Changes in electromyographic gait patterns of calf muscles with walking speed. IEEE Transactions on Biomedical Engineering, BME-30:73-76, 1981.

69. Simon, S.R., I.L. Paul, J. Mansour, M. Munro, P.J. Abernethy, and E.L. Radin. Peak dynamic force in human gait. J Biomech, 14(12):817-822, 1981.

70. Smidt, G.L. Hip motions and related factors in walking. Physical Therapy, 51:9-21, 1971.

71. Smith, K., C. McDermid, and F. Shideman. Analysis of temporal components of motion in human gait. Am J Physical Med, 39:142-151, 1960.

72. Soames, R.W. and R.P.S. Richardson. Stride length and cadence: Their influence on ground reaction forces during gait. In: Biomechanics IX-A, D.A. Winter, R.W. Norman, R.P. Wells, K.C. Hayes, and A.E. Patla (Eds.). Human Kinetics Publishers, Champaign, IL, 1983, pp. 451-456.

73. Soule, R.G. and R.F. Goldman. Energy cost of loads carried on the head, hands or foot. J Appl Physiol, 27:687-690, 1969.

74. Soule, R.G., Pandolf, K.B., Goldman, R.F. Energy expenditure of heavy load carriage. Ergonomics, 21:373-381, 1978.

75. Stulen, F.B. and C.J. De Luca. Muscle fatigue monitor: A noninvasive device for observing localized muscular fatigue. IEEE Transactions on Biomedical Engineering BME-29:760-768, 1982.

76. Sutherland, D.H. An electromyographic study of the plantar flexors of the ankle in normal walking on the level. J Bone Joint Surg, 48A:66-71, 1966.

77. Sutherland, D.H., R. Olshen, L. Cooper, M. Wyatt, J. Leach, S. Mubarak, and P. Schultz. The pathomechanics of gait in Duchenn muscular dystrophy. Developmental Med and Child Neurology, 23:3-22, 1981.

78. Winsmann, F.R. and R.F. Goldman. Methods for evaluation of load carriage systems. USARIEM Technical Report T40/76, 1976.

79. Winter, D.A., A.O. Quanbury, and G.D. Reimer. Instantaneous energy and power flow in normal human gait. In: Biomechanics V-A, P. Komi (Ed.). University Park Press, Baltimore, MD, 1976, pp. 334-339.

80. Winter, D.A. Biomechanics of Human Movement. John Wiley & Sons, Inc., New York, NY, 1979.

81. Winter, D.A. Moments of force and mechanical power in jogging. J Biomech, 16(1): 91-97, 1983.

82. Winter, D.A. Biomechanical motor patterns in normal walking. J Motor Behavior, 15(4): 302-330, 1983.

83. Yamashita, T. and R. Katoh. Moving pattern of point of application of vertical resultant force during level walking. J Biomech, 9:93-99, 1976.

84. Zipp, P. Recommendations for the standardization of lead positions in surface electromyography. Eur J Physiol, 50:41-54, 1982.

Appendix A

Nomogram used for sample size estimation (14)

For both one and two-tailed tests, the nomogram relates sample size, inter-trial correlation coefficient, and the smallest significant (p<0.05) mean difference between conditions, in standardized normal scores, that the experimenter wishes to detect. To use the nomogram, a straight edge is placed against both at the inter-trial r-value and the mean difference in standardized normal scores. The straight edge then crosses the left-side scale indicating the number of subjects needed.